THE EFFECT OF ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION ON LEG-SPRING STIFFNESS DURING HOPPING

David K. Wolfe

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Doctoral Committee

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Rafael Bahamonde, PhD Chair

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Jefferson Streepey, PhD

November 13, 2020

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Zachary Riley, PhD

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Kelly Naugle, PhD

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Matthew Beekley, PhD, CSCS

Acknowledgements

Working toward my Ph.D. in Health and Rehabilitation Science has been a long and laborious journey. It has meant a lot of hard work, and sometimes a lot of stress to get things done. None of that could have happened without the love and support of my wife Becky and our daughter Maryjane. Through the good and the bad times, they have always been supportive and encouraging of my progress, no matter the strain. In addition to my wife and daughter, the rest of our family has provided us with no end of emotional support and encouragement to preserver in both the ups and the downs. Without them, I wouldn't be here now.

This journey started with a conversation with Connie Pumpelly MS, ATC in the fall of 2009, who very frankly told me "if this is what you really want to do, and from the way I've watched you interact with the students it is, you really need to get your Ph.D." I had my reservations about starting coursework again, but as I was looking at potential programs, Becky actually first stumbled on the Health and Rehabilitation Science Ph.D. program, and brought it to my attention. I thought immediately that it was a perfect fit.

From there, other friends, family, and colleagues continued to provide guidance and encouragement as I began my journey. Ned Shannon MS, ATC, my supervisor at the time, was willing to work with me and my course schedule so I could attend classes, while others helped me with my transition back to being a student. Robert Aaron Ph.D., one of the many members of our church family, provided a great deal of encouragement, as well as helping me to stay grounded, while his wife Suzanne Aaron MA, provided advice to Becky on what it is like to be the wife of a Ph.D. student. More

than once, the Aarons probably saved our marriage; there are no words that adequately express my gratitude to them for that.

While there are too many names to mention, let it be said that the members of our church family at the North United Methodist Church have been a constant and positive presence in this journey. They have been there to provide encouragement, motivation, and to give me the occasional kick in the pants that was needed to keep moving forward, both figuratively (Tom Welch) and literally (Jeff Eggert). By the time this is in print, I'll have completed a half-marathon, something that I would have never dreamed of if it hadn't been for my research and the folks that are part of our church family.

Shortly after enrolling in the program, my position at the University of Indianapolis changed, and I became a member of the faculty in the Biology department. I partially credit Megan Palmer Ph.D. with my ability to make the move from Athletic Training to Biology, due to her stressing the importance of our (as Ph.D. students) ability to write an appropriate philosophy of teaching statement. Dr. Palmer, as well as the rest of my professors, have always been encouraging and supportive of my progress, for which I am very grateful. Moving to the Biology department was a fantastic opportunity at the time, for many reasons. Stephen Nawrocki Ph.D. was assigned as my faculty mentor, whose guidance and encouragement have been invaluable. John Langdon Ph.D. also became a colleague in my new position, and if anyone has pushed me to realize what it is to be a member of the faculty at the university level, it's Dr. Langdon.

Getting the keys to my office in the Biology department, I was immediately challenged by another colleague, Krista Latham Ph.D. She handed me my office keys (she was getting a new office, and I was to take her old one) and said, “You have to be the one. Several folks have been in this job while working on their Ph.D., and they all are now stuck at ABD. You have to be the one to get it done.” That statement, along with the persistent encouragement from my former undergraduate advisor, Dr. Pearlmarie Goddard, has kept me going many days.

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The digitizing and analysis of the data for this project pushed me in a completely different direction than I was expecting, and that was into the field of computer programing. Because of this, there are a few other folks that deserve special recognition for all of the aid and assistance they provided me in this process. One is a member of the UIndy community, Stephen Spicklemier Ph.D., the Physics department chair. When I started working on the computer program that I would need to analyzed my data, the word “Python” kept coming up. This was particularly vexing because I kept including the title MATLAB in my searches. It turns out that Python is another computer programing language, and that several folks in biomechanics (and other fields) use Python to analyze their data. I had little idea what I was getting myself into, but with

Steve's help, I went from downloading Python to having a working computer program that would start to analyze my data in approximately three months. Steve let me badger him with questions about everything from syntax to various commands, all the while saying that this was a chance for him to have "fun" with programming. What was a pleasant distraction for him was for me an absolute lifesaver, and I greatly appreciate it.

The other group that I would be sorely remiss if I did not recognize would be the organizers and members of the IndyPy Meetup group. At their monthly meetings, they not only tolerated my continual badgering with beginner-level questions, but they never refused to offer any assistance they could, regarding my project. Often, I would have one person answer one question that allowed me to make progress, and then someone else would come over and provide a solution to the next problem, many times before it had even shown up. This group also pointed me in the direction of other groups and resources, without which, I would never have completed the project.

Towards the end of my journey, some of my support structure changed, as I went from working at the University of Indianapolis to working as the Sugar Creek District Executive for the Crossroads of America Council, Boy Scouts of America.

Joining the professional scouting staff at Crossroads of America Council allowed me to focus on one of my favorite parts of teaching, the building of relationships with others in the community in central Indiana. This change in position also allowed me to teach in a different way, and to impact the lives of young people prior to getting into the college classroom. Equipping our young people (and their families) with life and

leadership skills through access to the scouting programs will ultimately build better community members and improve many more peoples' quality of life. That, in my opinion, is a good thing.

Other folks that have helped to make this possible are the members of my committee and the School of Health and Rehabilitation Science. Rafael Bahamonde Ph.D. as my key advisor has always found time to listen to my concerns and provide some advice and guidance, as well as helping me decide on my project. Joyce MacKinnon EDD, and more recently Brent Arnold Ph.D., have been present, pushing me to decide on a project, and keeping me on track with my courses. The technical support provided by Jefferson Streepey Ph.D., Zachary Riley Ph.D., Kelly Naugle Ph.D., and Matthew Beekley Ph.D. have also been of great help.

Earning my Ph.D. has not been a journey I have taken alone; I have had many fantastic people helping me all along the way. Without the help of the people mentioned above, as well as countless unnamed others, this would not have been possible.

Preface

I've been curious about the world around me for a long time, particularly as it pertains to watching people walk and move. Watching someone dance, run, walk or swim; it was all interesting and often gave a little insight into who they were as a person. As an undergraduate student majoring in Sports Medicine, I gravitated to the preceptors who did work with orthotics and the biomechanics of walking and running. At the time, I thought there was something almost magical about what could be done with some tape and a little bit of foam to help an athlete compete more effectively and to be pain-free.

Continuing my education with my master's degree in Physiology and Biophysics, I gained more knowledge and understanding about how the human body worked and was able to incorporate this into my interest in human motion. The functions of the cardiovascular and musculoskeletal systems are at the center of human motion, so one field naturally flowed into the others, forming a more comprehensive view of the factors involved in human motion. But I didn't stop there.

Even though much of the medical field is moving toward "evidence-based practice" there is still a blending of art and science in the practice of medicine. The building of orthotic insoles for therapeutic purposes is one of these areas, as is the field of physical medicine. The concepts of physical medicine (osteopathic manipulative medicine, physical medicine in rehabilitation, instrument assisted soft tissue mobilization, etc.) involves a more in-depth understanding of human anatomy and physiology, as well as how motion affect and is affected by anatomic structures and physiological activity.   They also defy categorization strictly into the realm of either "art" or "science".

Thankfully, some biomechanical activity is quantifiable and therefore can be studied and evaluated through scientific methods. Stride length, stride frequency, stance width, ground reaction force, joint angle, and many other factors can be easily quantified and used in calculations to determine many aspects of the gait cycle. Leg-Spring Stiffness is one of those calculations, giving a measurement of how well the lower extremity is doing regarding the absorption of shock due to ambulation. Leg-Spring Stiffness also reflects the function of the musculoskeletal and its ability to attenuate the forces involved in the gait cycle.

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Leg-Spring Stiffness (LSS) is the measure of the musculoskeletal, neuromuscular, and biomechanical functions of the human body, and an appropriate evaluation metric for changes brought on by Anterior Cruciate Ligament Reconstruction (ACLr). ACLr can lead to flexion and extension loss, resulting in increased stiffness of the musculotendinous units of the ACLr leg and thus changes in LSS. LSS can be measured using K_{leg}, but little is known about the validity and reliability of the different methods of LSS and K_{leg} calculations. The purpose of this study was to determine if ACLr leads to a change in LSS (as measured by K_{leg}) during hopping, and to compare results of the Spring-Mass calculation and knee Joint Torsional stiffness methods in the computation of the overall K_{leg}. Video data synchronize with GRF were used to compute the kinematic and kinetic variables.

Mann-Whitney U tests were used to determine significant differences between the control and experimental group for the Spring-Mass method of calculation ($p = 0.004$), Joint Torsional method ($p =0.44$), K_{knee} ($p = 0.29$), and K_{ankle} ($p = 0.17$). Cohen's effect calculations showed small to medium effects for the K_{Knee}, ($d = 0.383$) but moderate effect size for the K_{Ankle}, ($d = 0.541$). Wilcoxon Signed Rank comparison for all the legs and ($N=42$) between computational methods were significant differences between computational methods ($Z = 5.65$, $p = 0.000$), and with a large effect size (Cohen's $d = 3.14$). Similar results were found when comparing only the ACLr leg values

(p = 0.005, Cohen's d = 4.88). The comparison between ACL Leg vs Non-ACL leg for experimental group subjects was not significant in either calculation method (Spring-Mass p = 0.20, Z = -1.27; torque calculation p = 0.96, Z = -0.05).

The spring-mass method was more stable and able to detect differences between the control and ACLr group. The lack of statistical differences in the joint torsion calculation method, as well as in comparing the unaffected leg to the ACLr leg in the experimental group, suggests that LSS may not be a precise enough measurement to determine the effects of an ACLr.

Rafael Bahamonde, PhD, Chair

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Abbreviations

ACL – Anterior Cruciate Ligament
ACLr – Anterior Cruciate Ligament Reconstruction
ACLi – Anterior Cruciate Ligament Injury
ACLd – Anterior Cruciate Ligament Deficiency/Deficient
ADL – Activities of Daily Life
GRF – Ground Reaction Force
IDE – Integrated Development Environment
EMG – Electromyography
LSS – Leg-Spring Stiffness
MC – Motor Control
MCL – Medial Collateral Ligament
M1 – Primary Motor Cortex
S1 – Primary Sensory Cortex
TS – Triceps Surae (Gastrocnemius and Soleus Muscles)
PAR-Q – Physical Activity Readiness Questionnaire
PMC – Primary Motor Cortex
Froude number $= v^2/gl$ – Froude Number Equation
v^2 – Velocity (vertical)
g - Gravity
L – Leg length or hip height
$K_{leg} = F/\Delta L$ - Spring-Mass Leg-Spring Stiffness Equation
K_{leg} – Leg-Spring Stiffness
F – Vertical force, gravity
ΔL – Change in leg length
$K_{leg} = K_{knee} \div l^2 sin\Delta\theta$ - Torsion Leg-Spring Stiffness Equation
$K_{knee} = I(\Delta\omega^2 \div \Delta\theta^2)$ – Knee Joint Torsion Stiffness Equation
$K_{ankle} = I(\Delta\omega^2 \div \Delta\theta^2)$ – Ankle Joint Torsion Stiffness Equation
$d = (M\_2 - M\_1)/S\_pooled$ – Cohen's D Equation
$\Delta\omega^2$ – Change in Angular Velocity
$\Delta\theta^2$ – Change in Joint Angle
l – Length of Limb Segment

Chapter 1

Introduction

When running, hopping, jumping, and walking, the leg bends to absorb the forces of gravity and motion to help maintain a stable posture. A large component of this involves the flexion and extension of the knee joint of the lower extremity as part of a complex referred to as the Leg-Spring system. The musculoskeletal components of the lower extremity work together to allow as smooth and fluid motion of the center of mass of the body as possible while maintaining the appropriate activity requirements. Leg-Spring Stiffness (LSS) is the mathematical representation of these activities, and can be described as the "resistance to change in leg length after application of internal or external forces" [1]. LSS is measured in Newton-meters (Nm), and can change with variables such as training, gender, injury, and the compliance of the surface that is being used. Furthermore, any changes to the involved structures, such as the ligaments and tendons of the knee, as well as the muscles acting on the joints of the lower extremity, will influence how the Leg-Spring functions and how forces are attenuated.

Studying LSS involves understanding the "elaborate interactions between multiple physiological and biomechanical tissue properties, as well as between the musculoskeletal and neuromuscular systems" [2]. This also means that any changes to the above structures and components can have a wide range of effects on LSS and lower extremity function. With an eye to these factors, as well as to the increased rate of activity in the population, understanding LSS is an important consideration for the healthcare community. Given the premium that professional athletes place on sport performance, this understanding can also have a significant increase on athletic capabilities within sport competition. So understanding how LSS can be affected by an injury to the lower extremity would be a benefit to several aspects of society.

Second to an inversion ankle sprain and concussions [3], an injury to the Anterior Cruciate Ligament (ACL) is probably one of the most common injuries to competitive athletes resulting in a limitation to leg range of motion [4, 5]. According to the University of California at San Francisco Department of Orthopedic Surgery and Sports Medicine, as well as the American Orthopedic Surgeons Sports Medicine Association, there are approximately 200,000 Anterior Cruciate Ligament Injuries (ACLi) per year in the United States, with over 70% of them occurring in football, basketball, soccer, and skiing [3, 6, 7]. Of these injuries, approximately half to three quarters are surgically reconstructed. As high school, college, and amateur athletic competition becomes more popular, the frequency of these injuries is likely to increase.

The occurrence of an ACLi can happen as a result of extreme valgus stress due to contact with the knee, often resulting in the “unhappy triad” of an injury to the ACL, Medial Collateral Ligament (MCL), and the medial meniscus. An ACLi can also occur as a result of a rotational force across the knee when the knee is in a straight position; this is referred to as a non-contact ACLi. This can take place as a result of improper technique (a tackle across the body in soccer), a “plant and twist” type of mechanism where the foot is planted and the body is rotated externally, or where the foot is rotated externally in relation to a stationary body [8].

While the frequency of ACLi in the United States is approximately 1:1750 for the total population, when it is taken into consideration that these injuries occur predominantly in competitive and high-intensity athletic activities participated in by people between the ages of 12 and 45, this frequency is actually much higher: 1:38 in college athletes [3]. The frequency is also higher in women athletes when compared to men, approximately 1:11 [4, 7, 9-11], possibly due to the increased Q angle of the female leg, kinetic activity proximal to the knee joint, and altered firing rates of the quadriceps muscle group and soleus muscles [9-11], when compared to their male counterparts.

As sporting activities have become more popular, and people have become more active in their later years, the risk for ACLi has increased. Along with this increase, there have been substantial changes to the treatment of patients with ACLi. In the early and mid-1990's, medical consensus held that amateur athletes and the general population would undergo rehabilitation for an ACLi, along with bracing for any return to activity. At that point in time, ACL reconstruction surgery (ACLr) was reserved for competitive athletes at the college and professional level. It is now understood that an ACL deficient knee is one that will eventually suffer from arthritic complications due to the loss of stability provided by the ligament. To this end, it is now the standard treatment that an ACLr takes place for any injured patient [6] to prevent the repetitive anterior translation of the tibia on the femur. The ACL becomes tight when the knee is in the extended position, and the ACL, therefore, prevents the anterior translation of the tibial plateau on the distal end of the femur when walking and hopping. While the normal or native structure may be restored, the compensatory changes to the LSS mechanism remain unknown.

The LSS is affected by several different factors. The surface that is being moved across, whether or not footwear is being worn and the type of footwear, the speed and pattern of motion, the weight of the person, and any extra weight that they are carrying are just some of the things that affect LSS [12-20]. Combined with these factors, the sole of the foot ranks third on the primary somatosensory (S-1) cortex in terms of the amount of area devoted to sensory perception [21, 22]; only the face and hands rank higher. A combination of tactile and proprioceptive sensory inputs - from the sole of the foot and the muscles, tendons, and ligaments of the leg, respectively - are constantly analyzed by the brain's motor control centers with each movement. This motor control analysis results in the adjustment of LSS to adequately dampen the vertical movement of the center of mass while maintaining the propulsive forces required for body motion. As the stabilizer of the knee in extension, the role of the ACL becomes evident. The loss of proprioceptive feedback from the ACL [23-25] and changes in the Leg-Spring system mechanism may effect LSS. This loss of proprioceptive feedback may also result in compensation by the other intact/non-injured structures of the knee and leg in question, as well as compensatory mechanisms in the unaffected leg [23-29].

As such, any change in LSS requires the interaction of all the structural components of the leg to resist the forces that are applied [1]. The purpose of this study was to determine if subjects with ACLr have a change in LSS during hopping, as measured by the biomechanical methods of Spring-Mass and joint torsional K_{Leg} calculation.

Statement of the problem

LSS is calculated in several ways, and from a variety of data types. Each of these methods can provide different types of information about the function of the lower extremity [1, 30, 31]. LSS has also been studied in regard to a host of different medical conditions, and in several different populations. Rarely, if ever, has LSS been studied in a manner that would allow it to be used as a diagnostic tool - pointing to the presence of a particular pathology. Patients with an ACLi and subsequent ACLr will have some decrease in range of motion (ROM) of the leg at the knee [4, 32-39], as well as changes in muscular stiffness of the hamstring and quadriceps muscle groups [5, 40, 41]. As the frequency of these injuries increases [4], so does the occurrence of this deficiency in the active population. Due to the prevalence of these injuries and their potential effects on hopping, it has been prudent practice to study the effects of an ACLi on neuromuscular function and force production of the knee [32, 33, 35, 38, 39], alterations to the knee ROM [34, 36], and loss of proprioceptive control [25, 42]. The loss of neuromuscular function from the injured and repaired ACL will lead to an alteration in knee joint performance. This alteration will directly affect the mechanics of the Leg-Spring.

Purpose and Significance of the Study

The purpose of this study was to determine if and how LSS in hopping was affected by the ACLr procedure, and to compare those effects on the two methods of LSS calculation. The specific aims were:

1. To determine if ACLr subjects had a quantifiable change in LSS during hopping compared to the control (non-ACL) subjects, as measured by changes K_{leg} Spring-Mass, K_{leg} Joint Torsion, knee and ankle stiffness (K_{Knee} and K_{Ankle}, respectively).
2. To determine if there are differences in LSS calculation methods, when comparing vertical center of mass displacement (Spring-Mass) with knee torsion stiffness calculation methods, as well as between the control/non-ACL and experimental/ACL groups.
3. To determine if ACLr subjects had a quantifiable change in LSS in the affected leg as opposed to the unaffected leg, within the experimental group, as measured by changes in knee and/or ankle stiffness (K_{Knee} and K_{Ankle}, respectively).

These aims were achieved through biomechanical evaluation of subjects between the ages of 18 and 25 - who have had an ACLr and completed a full course of appropriate rehabilitation treatment. The extent of this rehabilitation treatment was determined by verbal verification that the subjects were cleared to participate in athletic competition or had at least been cleared for normal activity upon retiring from organized high school or college athletic competition. The time span from medical clearance to study participation ranged from a low of 8 months post ACLr to a maximum of 6 years.

This study will add to a body of work relating biomechanical anomalies resulting from the injury, repair, and rehabilitation of the lower extremity. This study will also become part of the field of data that relates to “real world” aspects of human motion, and how LSS can be quantified and changed over time, as a result of an injury, and how rehabilitation techniques may be altered or improved.

Scope of Study/Limitations of Study

The data from this study will be beneficial to the medical community, particularly those working in the rehabilitation setting, by helping to refine what methods are most appropriate for measuring changes in LSS. It will also lead to the possibility of LSS data being used in a diagnostic role, signaling that a pathology may be present. Furthermore, those fields that deal with the increasingly active sections of society will benefit from this research. Additional information on how LSS is effected by ACLr will be critical to the advancement of treatment of active and athletic populations.

To be included in this study, participants (ACLr and control) must have been between 18 and 25 years of age, with normal abilities to walk and hop. ACLr participants must have had their patellar tendon graft ACLr at least 12 months prior to enrolling in the study and be cleared for normal activities. Participants must also be free of confounding medical conditions, such as medial and lateral meniscus tears, and damage to the other knee ligaments.

While the use of biomechanical analysis has been used in the field of ACL rehabilitation before and will continue to be used, it is a labor-intensive process, so its use is primarily confined to research studies. A point of error in the calculations may arise from the concept of “repetition without repetition”[43], where the end goal of the hopping task may be the same, but the actions of each joint with each hop may be different. Further limitations come from the recognition and digitization of the markers themselves. All of this can lead to some variance in the results of biomechanical analysis. The absence of matched groups between the control and experimental groups may call the validity of the statistical significance into question, as may the composition of the groups themselves. The control group is composed mainly of a sample of convenience, while the experimental group is almost completely composed of Division 1 and Division 2 college athletes.

Summary

LSS is a widely used evaluation tool for understanding the actions of the lower extremity. ACLi is one of the more common injuries, particularly in athletes on all levels, as well as those with active lifestyles. The frequency of these injuries has increased in recent decades, and with that, so has the surgical repair of ACLi become the norm. Even with the increase in surgical repair, there are likely to be changes to the biomechanical function of the lower extremity once the repair and rehabilitation have been completed. These changes are likely attributable to a loss of proprioceptive inputs from the damaged ACL, leading to changes in LSS. This study will provide new insight into the field of LSS and AClr rehabilitation as it addresses the fact that there has been little research conducted on how LSS when hopping, is affected by an ACLr.

Chapter 2

Literature Review

Introduction

The study of the LSS and the biomechanics of the hopping ACLr patient requires an understanding of several fields and the ability to bring them all into perspective. This injury and its subsequent repair and rehabilitation can be evaluated with biomechanical analysis because it exerts a specific pathologic effect: altering the physiologic capabilities of the tissues in the lower extremity. To maintain mobility in the face of these changes to normal physiologic function, there are alterations of the motor control pathways to compensate for the loss of proprioceptive feedback provided by the nervous receptors on the ACL. These adaptations are driven by both normal activities of daily living, as well as the exercise activities that are a staple part of the ACLr rehabilitation process.

Leg-Spring Stiffness

LSS is a measure of the musculoskeletal and neuromuscular response to a stimuli. It is the continual balance between the mobility required for a particular task of the lower extremities, and the stability necessary for safe and efficient motion [1, 2, 43-47]. The elastic components and the physical arrangement of muscular fibers of the glutes, quads, triceps surae (TS), and other muscles of the leg allow for spring energy to be stored as the body moves through space and time [2, 48-56]. Other factors at play are the joints of the lower extremity, as well as the degrees of freedom of each joint that is involved [43, 46, 47]. These variations in LSS are also due to Bernstein's hypothesis of “repetition without repetition”, where a movement can be consistent with its end result, but different in its component parts and movements [57-59].

The stiffness of this spring adjusts automatically through the feedback mechanisms associated with the musculotendinous and ligamentous structures of the lower extremity, resulting in minimal displacement of the center of mass [15, 16, 60-64]. As the surface that activities are performed on becomes more compliant, the Leg-Spring will become stiffer, while as the surface becomes stiffer, the Leg-Spring will become more compliant [12, 15, 17, 19, 65, 66]. This is due to the components of the lower extremity automatically adjusting to the conditions presented while acting as torsional springs, whose compression and energy storage is manipulated through the change in angular motion and angular velocity as impact forces from the ground are dealt with [67, 68]. The sum of the total of these changes in the angular activity of the leg joints is what is reported as the measure of LSS.

LSS itself is the sum of the total of the contributions of the musculoskeletal components of the lower extremity dealing with impact forces due to motion and gravity. From the motions of the toes and the arch of the foot, up to the rotation of the various joints of the pelvis and sacrum on the lumbar spine, each component plays a role in adjusting to the surface being traversed. These alterations to LSS are due to the feedback from the muscles, tendons, and ligaments of the lower extremity, providing information that leads to LSS adjustment [23-27, 32, 69-71]. Hopping and running allow the foot to function as a 2nd class lever, with much of the force being applied by the TS muscles. The TS is the largest muscle group in the leg, and the motions of the foot at the ankle joint allow the TS muscles to make much better use of the elastic and energy-storing components of the leg portion of the lower extremity [72, 73].

Since the Leg-Spring deals with the movement and attenuation of the force of a human body moving across a surface, it also provides a useful measure for the effectiveness of several different types of equipment. From braces that support and protect the knee and ankle, the presence or absence of various types of footwear, speed of movement, to the type of surface used to make a running track or activity room floor, measuring LSS can provide a great deal of information about how the body will function under various conditions [15, 44, 51, 63, 66, 74-80].

The surface encountered, whether it is being hopped, ran or walked on, will have a great deal to do with the stiffness of the leg spring. The more compliant the surface, the greater the LSS [15, 63, 77]. As was also shown in several studies, the speed across the surface also increases the stiffness of the leg spring mechanism. The presence or absence of shoes, as well as their construction, will also play a similar role in the action of the leg spring [51, 66, 75, 76]. Barefoot running leads to decreases in LSS, due to the need of the joints to flex to absorb the downward acceleration of body mass, while simultaneously store elastic energy in the elements making up the leg spring. This allows this stored energy to be returned in the later stages of the gait cycle.

Gender and frequency of hopping also play a significant role in the LSS measurement [11, 15, 16, 20, 41, 61, 63, 78, 81-83]. Q angle is the angle of two lines drawn on the thigh, with one line running from the anterior-superior iliac spine to the tibial tuberosity (the insertion point of the quadriceps muscles on the tibia), and the other line running down the center of the long axis of the femur, also through the tibial tuberosity. In females, this angle is significantly larger than in males, due to the wider set of the pelvic structures that are necessitated for childbearing. Due to their larger Q angle, females have a lower LSS than their male counterparts [11, 81, 82]. The lower LSS in females is a compensatory mechanism that allows for maintenance of a bipedal posture, with support structures remaining under the center of mass, despite the larger Q angle. With injury or increased hopping frequency, this discrepancy in LSS, based on gender, decreases, with the increase in female LSS eventually meeting that of their male counterparts [10, 41].

As the hopping frequency increases, LSS also increases in both males and females, with the difference between male LSS and female LSS also decreasing as speed or hopping frequency increases [61, 63]. This increase in LSS is due to the increased need for efficiency of motion, as well as the increase in the protective nature of muscular activity. As speed increases, so do the risks for injury due to excessive motion around a joint. So as the speed or frequency of hopping increases, joint motion will decrease to provide the needed stability. This decrease in joint motion will increase LSS accordingly [61, 63, 67, 84]. When hopping frequency is self-selected, as it was in this study, it settles in to a fairly consistent 2.2Hz across all populations, regardless of training level or gender [61, 63]. This frequency provides the necessary stability for the joints of the lower extremity to perform the hopping activity with relative efficiency and endurance, as well as a consistent involvement of the control structures.

Multiple papers authored by Farley, et al [15, 63] have dealt extensively with the adjustment of LSS in regard to changes in speed and surface. These studies, one of which has been corroborated by Arampatzia, et al [61], show the increase in LSS with the increase in running speed. These studies show that with increases in speed, the degree of flexion of the lower extremity joints decreases just prior to the point of surface contact, thereby increasing the LSS upon contact with the ground. This same mechanism is at play with respect to the compliance of the surface – a more compliant surface will result in less flexion of the joints upon contact, increasing the LSS. So while there is an instantons change in LSS brought on by a change in surface, hopping frequency, or running speed, there are also long term adaptations to LSS that will further enhance those changes.

These studies by Farley, et al, and Arampatzia, et al, also emphasize the need to look at individual joints, particularly those further down the kinetic chain, when trying to decipher where LSS adjustments are taking place [15, 61, 63]. The ankle joint plays a critical role in the adaptation of LSS to changes in speed and surface compliance. Since the foot and ankle are the first move upon touchdown contact, they play the most immediate part in decelerating the rest of the body in the vertical direction, as well as being first to store elastic strain energy in the muscles of the leg [78].

Elite athletes, regardless of gender, have higher LSS than their untrained counterparts [85]. This is due to increased training leading to increases in economy. Increased LSS leads to more storage of elastic strain energy, as well as a decreased oscillation of the center of mass [61, 86]. Untrained individuals not only have a lower

LSS, they also have greater vertical oscillations of the center of mass, as much more energy is wasted in vertical, instead of horizontal, motion.

Injury can also increase LSS due to alteration of the normal anatomical structure, as well as the decreased amount of feedback from the neuromuscular structures of the leg [23-27, 32, 69-71, 87-90]. Studies that looked specifically at the ACLd and ACLr knee cited increases in bilateral LSS as a neuromuscular adaptation to the injury [32, 91]. Additionally, the decrease in control information available for neuromuscular coordination leads to a decrease in the available degrees of freedom in the joints of the lower extremity throughout the range of motion [32, 47, 70, 71, 87].

Motor Control and Physiology

Under normal conditions, there is considerable motor control (MC) of the lower extremity allowing for the maintenance of balance on the lower extremity when at bipedal rest, as well as while controlling the motion of the lower extremity during activity. All bipedal activities require the actions of hip flexors and extensors, hip internal and external rotators, quadriceps, hamstrings, triceps surae (foot plantar flexors), foot dorsiflexors, and intrinsic muscles of the feet to be coordinated in an organized fashion. All control commands from the primary motor cortex (PMC, also known as the M1) must first be coordinated with proprioceptive information from the cerebellum and the conscious MC activity in the premotor cortex. These MC commands are also shaped by visual, vestibular, and reflexive inputs from their respective systems. Once these control commands are integrated, the M1 can activate the appropriate muscle groups in the correct sequence and activity level to accomplish the motion in the lower extremity.

Following ACLi, one of the compensatory mechanisms involves the alteration in muscle firing order. Changes to the motor control systems include a “feed-forward” type of neuromuscular adaptation, leading to the anticipatory activation (pre-activation) or coordinated co-activation of the knee musculature to protect the knee joint structures in the presence of an ACLd [41, 92, 93]. This allows for the prevention of excessive anterior translation of the tibia on the femoral condyles, resulting in the dynamic stabilization of the knee [5, 40, 41]. While injury to the ACL and its subsequent repair can restore a close approximation of native anatomic structure, this injury results in the loss of sensory inputs that came from the intact ACL [23-25, 69-71, 94]. Because of the loss of these proprioceptive inputs, studies dealing with the kinematic alterations due to ACLr have suggested that part of the gait alteration is due to a decrease or alteration in neuromuscular control [26, 32, 35, 93, 95].

In their article dealing with dynamic stability of hopping ACLd subjects, Rudolph, et al [91] stated that subjects with successful musculoskeletal stabilization of the knee joint were able to hop normally. These successful subjects were referred to as “copers”, while those that did not have a successful stabilization of the knee joint were referred to as “noncopers”. Rudolph, et al were also able to show that there was bilateral symmetry between the effected and unaffected sides of the body in the coper study participants. These findings were corroborated by Bryant, et al [32] in their study which looked at the dynamic stability of ACLr subjects, and also showed an increase in bilateral stiffness was advantageous for lower extremity dynamic activity.

Because the goal of ACLr is to restore the native dynamic stability of the knee joint, the insertion of the graft tissue should always be accompanied by appropriate rehabilitative exercise. This rehabilitative exercise is important to restore the appropriate musculature balance between the anterior and posterior compartment muscles of the leg and thigh. Even when muscular strength and muscular balance has returned to “normal”, there can still be deficiencies that exist in the biomechanical chain, and those deficiencies become part of the “new normal” for the patient [26]. This is because of the loss of motor control inputs associated with both the ACLi and ACLr [26, 96-98]. It is these alterations to normal gait and knee activity that are thought to be observable through biomechanical gait analysis.

Methods of Calculation

There are several methods of LSS calculation, each of which can be based on a different set of kinematic and force data points [1, 30, 31]. The spring-mass model, equation 2, uses the ground reaction force (vertical force, GRF) and the change in leg-length to calculate the compliance of the leg spring. This method has been used to great success in the evaluation of different types of footwear, as well as the evaluation of the compliance of different surfaces [15-17, 19, 20, 63, 78, 99]. There is also considerable literature demonstrating this calculation method usefulness in subjects with an ACLr [4, 5, 8, 27, 28, 32, 33, 36, 39, 40, 61, 89-91, 98, 100-105]. The spring-mass calculation is also the most common method of calculating LSS [1, 15, 16, 31, 61, 63, 76, 106-108]. A similar, and also common method of calculation is to compute LSS as a function of GRF and the change in the displacement of the center of mass [1, 17, 19, 61, 63, 68, 76, 106-109]. While this method also utilizes GRF, it will result in different values for LSS (measured and reported as K_{Leg}), due to the incorporation of the movement of the center of mass in the air, as well as while the Leg-Spring is compressing and extending, and is in fact reporting vertical stiffness, not LSS.

The advantage of these two calculation methods, mentioned above, and those that are similar to them, is that they both divide the GRF by the change in distance between two points, making them both easy to calculate. This is advantageous because any two points in three-dimensional space are always on the same line and in the same plane. It does not matter if one of those point is the center of mass or the iliac crest, and the other is the floor, force plate, or distal most point of the foot. The change in length will always be observable as a change in length of a two-dimensional line. This makes both the calculation of LSS, as well as the understanding of what LSS represents, fairly straightforward. In these calculation methods, the resultant LSS value is a function of the actions of all the musculoskeletal components in the limb performing the task. The actions of each of these components can vary from repetition to repetition, but the result will be consistent [57, 58]. Also, in a study that reviewed multiple methods of LSS calculation, the studies that used the spring-mass method were considered more accurate than other methods, due to the ease in accurate marker placement on the bony prominences of the hip joint [1], the two dimensional nature of the points being measured, and also being more consistent across a variety of hopping frequencies [30].

The downsides to these methods of calculation include that they do not provide insight into the motions of individual joints of the lower extremity. A pathologic condition, musculoskeletal injury, the level of training, the subject's gender, or change in surface being moved across can all provide changes in the resultant LSS values [1, 11, 13-17, 19, 20, 30, 61, 63, 65, 75, 78, 79, 81, 82, 85, 99, 110]. While the LSS of the subject will be affected, there is no way of knowing from those calculation methods which joint or joints are most affected, or to what extent those effects apply. These methods of calculation also rely on a forceplate to collect GRF data. The use of a forceplate requires dedicated space and electronics, often costing well over $20,000. These costs can be prohibitively expensive for the clinical setting or at a teaching university.

As stated previously, LSS is a neuromuscular and musculoskeletal response to the activities of human motion, and can be calculated by using joint torques and anthropomorphic measurements, like the ones shown in equations 3, 4, and 5. These measurements can yield similar results to the spring-mass method [31, 44, 68, 80], but the values of each method of calculation are not interchangeable [1, 30]. The joint torque calculations are based on two-dimensional kinematic parameters representing the motions of the knee joint. This allows for an examination of a single joint and that joint's effect on LSS. There is a further benefit of the simplicity of data collection. Video data is all that is required for this type of calculation, with the frame rate of the camera being used to calculate angular frequencies and velocities as needed. The anatomical landmarks involved in the calculations can be taken directly from the video, plotted as coordinate data, and the vectors and motions calculated accordingly. This makes the calculations fairly straightforward, but it also introduces a degree of error into the calculation of LSS.

The basic rules of geometry state that any two points form a line, any three points form a plane, and any more than that must be dealt with in three-dimensional space. When looking at the multiple specific anatomic landmarks used to calculate LSS in this manor, if the multiple degrees of freedom of each joint are not taken into account, the result will be an inherent level of inaccuracy [43, 46, 47, 57, 58]. Another problem with this type of calculation is that it assumes that the action of each joint in a repeated task, such as hopping, will be identical with the previous and following hops [57, 58].

The torque method of calculation has not been studied in ACLr subjects, and does not require a force plate. The torque method of calculation also uses two-dimensional kinematic data derived from a high-speed camera. While the ability to take these images has become easier in recent years due to the presence of this capability in most cell phone cameras, this method's difficulty lies in the fact that it is much more mathematically intense. This method also relies on the accurate measurement of knee joint moments of inertia, which is more accurately performed with three-dimensional kinematic data [1], instead of the two-dimensional data that is more commonly used [1, 31, 107, 111]. This three-dimensional kinematic data is much more appropriate, due to the knee working in sagittal, transverse, and frontal planes, and two-dimensional kinematic data ignores the rotational components of the leg and thigh, as well as the varus and valgus stresses [1, 28, 29, 69, 87, 89, 90, 95, 101].

The study that the torque method of calculation was taken from, written by Dutto and Braun [31], was chosen for some fairly specific reasons. The torque method does not require the use of forceplates to collect GRF data. This allowed for LSS calculations that were based on two-dimensional kinematic data only, which could be captured by a basic video system and did not require a dedicated lab space. In this article, the calculation methods also had results that were comparable to those of the spring-mass model of LSS calculation. While this study has been cited many times for its findings on exercise induced muscle damage, its torque method of calculation was only cited in articles comparing methods of LSS calculation [1]. As the study progressed and the data was analyzed, and other studies were reviewed, a discrepancy between the calculated values in the study by Dutto and Braun and the results of this study began to appear. While joint stiffness values were consistent with what Dutto and Braun found, LSS values were consistently off by a considerable amount. Eventually, it was found that the actual value of the discrepancy between this study and that of Dutto and Braun was a factor of approximately 57.3, or the conversion factor necessary to convert from radians to degrees. Applying this variable to the torque calculated LSS values, they came in line with those reported by Dutto and Braun, but were no longer able to carry the Nm unit. The findings of this discrepancy have been corroborated by other studies, citing that the torque calculation method is consistent with itself, but not able to be compared to other methods of calculating LSS [1, 2, 30].

Regardless of motion or activity, method of calculation, or the presence or absence of footwear, the knee will flex to continue the absorption of ground reaction forces, attenuating and storing energy in the patellar tendon and quadriceps muscles. The elastic energy stored and the LSS itself will be dependent on the stiffness of the surface being covered, as well as the speed and gait style being used (hopping, running, walking, etc.). The result of this is a balance of the vertical and horizontal displacement of the center of mass of the body [60, 63, 67, 112-115].

Keeping the torso upright, the gluteal muscles eccentrically contract as the hip flexes and the Leg-Spring compresses with each foot impact with the ground. While the flexion of the hip is minimal, it is still noteworthy in the discussion of LSS, because the hip motion is necessary for maintaining the upright posture of the head and thorax. This helps to maintain the smooth and fluid motion of the center of mass and the head during walking, running, and hopping. With these Leg-Spring components combined, the range of elastic energy storage and recovery across the kinetic chain varies considerably, from well over 50% to almost none [67]. This range of activities, combined with the variety of methods of LSS calculation, means that a range of results can be expected, and must be considered in terms of their calculation method.

Biomechanics

"Running and hopping have been characterized as actions that enable an individual to move along the ground like a bouncing ball" [15, 67, 116]. The study of biomechanics as it relates to LSS is an attempt to put that bouncing motion into a language that allows an examination of the functions of the spring providing the bounce itself. Robert Hooke first put this into mathematical language in the 17th century with the law that bears his name. In Hooke's formula, the force is equal to a constant multiplied by the distance the spring is compressed or extended. With a simple bit of mathematical rearrangement, we get our spring-mass calculation for LSS (K_{leg}):

$$\boldsymbol{K_{Leg} = F \div \Delta L} \qquad \textbf{Equation 1}$$

With *(F)* being the maximum vertical force, and *(ΔL)* the change in leg length [15, 61, 67, 77, 117]. Maintaining the concept of the spring-mass equation as merely a reorganization of Hooke's Law, as shown in figure 1, the force and change in length are known, and the constant is what is being solved for. This allows for changes in the material properties of the spring, such as changes due to ACLr, surface, training, any change in vertical compliance due to muscular activity and proprioception to be accounted for in the reporting of LSS.

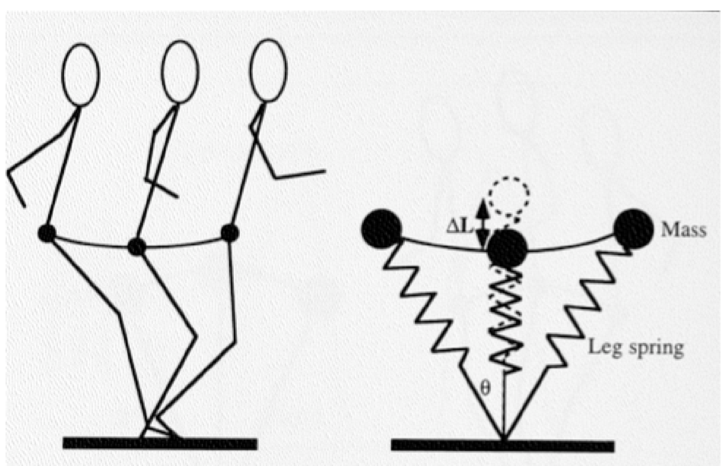

Figure 1. Leg-Spring-Mass Model [45]

LSS can be used as an evaluation tool of the entire lower extremity, due to "changes in leg geometry at foot strike alter the load torque about each joint" [67].

LSS can also be calculated by using the joint torsion stiffness using the following formula [31]:

$$\boldsymbol{K_{leg}} = \boldsymbol{K_{knee}} \div \boldsymbol{l^2 sin\Delta\theta} \qquad \textbf{Equation 2}$$

Where K_{leg} is LSS, l is the participant's thigh length, ΔΘ is the change in the angle of the knee joint (for equation 2), and K_{Knee} is the stiffness of the knee joint (as calculated in equation 3). The knee and ankle torsion stiffness can be calculated using the following equations:

$$\boldsymbol{K_{knee}} = \boldsymbol{I}(\Delta\boldsymbol{\omega^2} \div \Delta\boldsymbol{\theta^2}) \qquad \textbf{Equation 3}$$

$$\boldsymbol{K_{ankle}} = \boldsymbol{I}(\Delta\boldsymbol{\omega^2} \div \Delta\boldsymbol{\theta^2}) \qquad \textbf{Equation 4}$$

Where K_{joint} is the stiffness of the joint in question, ω is the knee or ankle angular velocity ($rad*s^{-1}$)(equations 3 and 4, respectively), Θ is the knee or ankle angular position in radians (equations 3 and 4, respectively), l is the length of the thigh or foot squared, multiplied by the participants mass (equations 3 and 4, respectively), and ΔΘ is the change in angle of the knee joint and ankle joint (equations 3 and 4, respectively) [1, 15, 31, 63, 118]. The length of the thigh or foot was measured using the markers placed to evaluate joint and leg movement, with the thigh measured between the knee and hip markers, while the foot was measured between the markers placed on the lateral calcaneus and the base of the 5th ray.

Summary

The measure of LSS has many components to be accounted for, from the motions of the individual joints, footwear and the surface that is being traveled. When an injury is present such and ACL, there are changes to the neuromuscular components of the leg. The protective mechanisms of the body, combined with the loss of native tissue and the lack of feedback from any graft, will all combine to establish a “new normal” for the range in which the Leg-Spring operates. This should be reflected in the overall LSS and the actions of any of the component joints involved in the calculation.

Chapter 3

Materials and Methods

Subjects

The hopping data collected included 21 subjects (18 – 25 years old): 11 controls, 10 subjects with ACL reconstructions (ACLr, see Table 1.). Subjects, both ACLr and controls, were recruited from Indiana University – Purdue University, Indianapolis and the University of Indianapolis.

Variable	ACL Group (N=10)	Control Group (N=11)
Gender	♀ = 7, ♂ = 3	♀ = 6, ♂ = 5
Height (cm)	173.6 ± 8.8	170.4 ± 10.2
Weight (Kg)	73.9 ± 12.5	74 ± 25.3

Table 1. Subject Anthropometrics

Enrollment in the study was dependent on the subject's successful completion of the Physical Activity Readiness Questionnaire (PAR-Q), the absence of any medical condition or surgery that would compromise the biomechanical nature of the study (ex: hip or ankle reconstructive surgery, etc.). Upon enrollment into the study, all individuals completed an informed consent form, advising them of the nature of the study and activities involved in data collection.

Of the subjects in the experimental group, six of the ACLr subjects had their injury and surgical repair on their right knee, while four had the same procedure done on their left knee. The most common form of ACLr, and the procedure specified for use in all subjects in this study, is a patellar tendon graft. Three of the experimental group subjects were male, while the remaining seven experimental group subjects were female.

As shown in Table 1, a total of eight males and 13 females participated in this study, with most of the experimental group being NCAA Division 1 or Division 2 college athletes.

Experimental Procedures

The experimental procedures included a two-hour data collection session for each individual, conducted in the biomechanics laboratory of the Department of Kinesiology at Indiana University – Purdue University, Indianapolis (IUPUI). Upon entry into the laboratory, study participants were enrolled in the study upon successfully completing the Physical Activity Readiness Questionnaire (PAR-Q) and IRB approved informed consent form.

Biomechanical markers were placed on at the base of the little toe (distal end of the 5th metatarsal), on the side of the heel (lateral border of the calcaneus), on the ankle bone (lateral malleolus), knee joint, and the point of the hip (midpoint of the iliac crest), on both legs (See Figure 2). While not standard biomechanical practice, the choice to place the marker on the lateral border of the calcaneus instead of the posterior border was made to facilitate more accurate measurement of the foot for the calculation of ankle torque, as shown in equation 4. At this point, all participants had the experimental procedure demonstrated for them. Following this, and the answering of any additional questions, data collection began with the study participants performing a

10-second hopping trial, on two AMTI (Advanced Mechanical Technology, Inc., Watertown, MA) force platforms. Once this 10-second trial was completed, the equipment was reset and the study participant would reverse direction for recording of the other leg's activity. The study participants were videotaped using two-dimensional (2D) motion capture via a Casio Exilim EX-FH20 video camera operating at 210 fps. ARGUS™ motion capture software was used to digitize the video data and to compute the 2D coordinates of the biomechanical markers, which was used to calculate other biomechanical parameters.

The ARGUS™ software was based on the original product built on the MatLab (MathWorks & Simulink Technologies, Natick, MA) programing language by Dr. Tyson Hedrick at UNC-Chapel Hill [119] and adapted to run on Python 2.7 programing language.

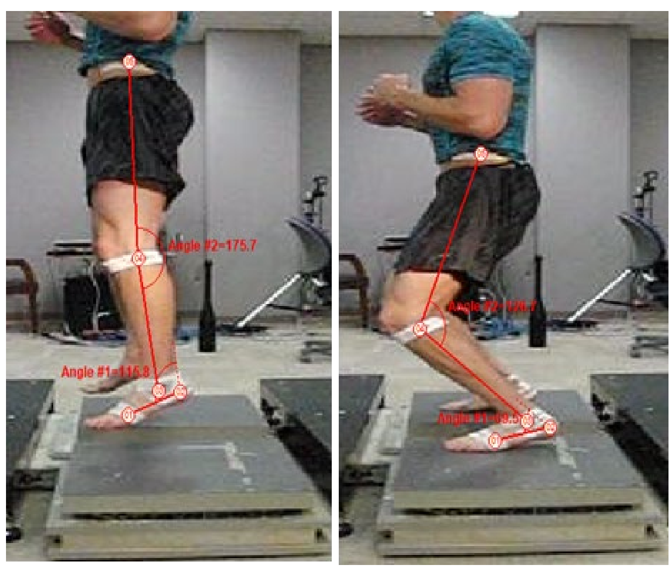

Figure 2. Joint Angles

As shown in figure 2, the two force plates are arranged parallel to each other, with the space between them aligned with the direction the subject was facing. When the hopping trials were conducted, one foot landed on each force plate, so the data reported for the calculations would just be for the testing leg. This means that the reported ground reaction forces are accurate for the motions and calculations for the tested leg, but represent roughly half that of the bodyweight of the subject hitting the force plate.

The data from both the control group and the experimental/ACLr group were inputted into the ARGUS™ software to generate a two-dimensional model of the movements of the ankle, knee, and hip joints while hopping. This process enabled calculation of the change in leg length during the weight-bearing phase of the hopping cycle. These data were paired with the ground reaction forces gathered as a result of foot strike and weight acceptance in the stance phase of the hopping gait cycle. This was accomplished by visual synchronization of the video frame at the point of ground contact with the starting of the recording of ground reaction forces. The maximum GRF from the force plate data, present at mid-foot strike, was combined with the leg length data to compute the LSS parameters.

Biomechanical Analysis

Each participant performed a double-leg hopping activity for approximately 10 seconds, at a self-selected speed. Once this was completed for one leg, they reversed direction and complete the hopping trial a second time for the other leg. All data from each trial, for each participant, was saved with a marker indicating either control or ACLr leg, with no personally identifying indicators. The coordinate data generated from both control and ACLr trials was used to generate a 2D joint angles of the lower extremities of the study participants, allowing the calculation of the variables needed to compute LSS.

Once video and GRF data collection had been completed, the ARGUS™ software was used to compile the biomechanical coordinate data for the calculation of LSS in purpose-built Python programs. Digitizing the video data involved advancing the video to the first frame of contact with the force plate. Then, each video was iterated five times, with one time for each of the biomechanical markers. This provided the 2D coordinates necessary for the calculation of the spring-mass and torque values of the human body in motion.

Two dimensional joint angles were calculated by taking the cosine of the absolute value of the angle created by the appropriate vectors (knee: knee-hip and knee-ankle; ankle: ankle-knee and ankle-head of 5th ray, see Figure 2). The absolute value of the angle was used to eliminate potential negative joint angles due to the direction of the coordinate data, see equation 6.

$$\boldsymbol{\Theta} = \left| \boldsymbol{cos}^{-1} \left(\frac{\boldsymbol{u} \cdot \boldsymbol{v}}{||\boldsymbol{u}||\,||\boldsymbol{v}||} \right) \right| \qquad \textbf{Equation 5}$$

In this equation, Θ is the angle of either the knee or ankle, and *u* and *v* are the unit vectors for the knee- hip and knee-ankle, or ankle-knee and ankle-head of 5th ray, respectively. The absolute value of the respective knee and ankle angles were taken to eliminate negative angles, due to the reversal in the direction of travel in the experimental procedure. These angles were used to determine the maximum and minimum joint angles for both the knee and the ankle for hopping trials, as well as maximum angular velocity. The maximum angular velocity was calculated by taking the maximum and minimum joint angles and dividing the change in angle by the change in time between the maximum and minimum, as the leg-spring was compressed (weight acceptance half of the gait cycle). While full extension of the knee is considered to be 180°, and the neutral position of the ankle is considered to be 90°, these measurements are only applicable to a person standing upright and flat-footed, with their knees "locked". As there is no set angle for either the knee or the ankle when not weight-bearing (the knee does not straighten to an exact 180° during dynamic activity, the ankle does not rest at 90° when non-weight bearing, etc.), the actual change in joint angle was used in the calculation of joint torques. This data was then used to compute ω, which is the knee or ankle angular velocity (rad*s^{-1}), and Θ, which is the knee or ankle angular position in radians, as described in equations 2, 3, and 4.

Computation of Leg-Spring Stiffness

Mid-foot strike and the initial weight acceptance phases of the hopping motion compress the Leg-Spring, with the rebounding force occurring at toe-off and the later stages of the hopping motion. As the Leg-Spring is compressed, many of the muscles of the lower extremity are contracting in an eccentric fashion while a stretching and lengthening force is being placed on the tendons. This attenuates the compressive forces due to motion and gravity while maintaining a vertical posture. The magnitude of LSS affects the vertical displacement of the center of mass, as well as the speed of motion and the duration of the hopping motion.

Upon ground contact, the leg acts as a spring, absorbing forces generated by gravity's effect on body weight, as well as the associated ground reaction forces and changes in kinetic energy due to forward motion (see Figure 3).

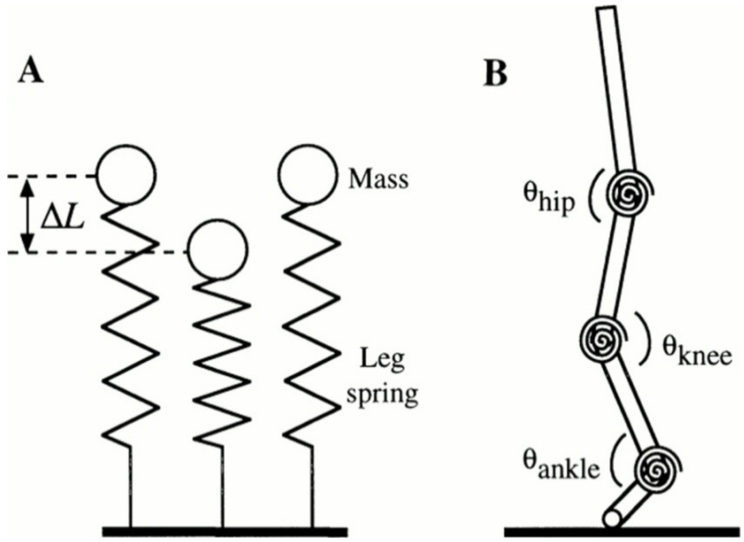

Figure 3. Leg-Spring [15]

Measuring LSS is a non-invasive procedure that involves the use of the mathematical equation (see Eq.3):

$$\boldsymbol{K_{leg} = Peak\ F_{g,y} \div \Delta L} \qquad \textbf{Equation 6}$$

Where (K_{leg}) is the stiffness of the Leg-Spring, Peak ($F_{g,y}$) is the maximal vertical component of the ground reaction force, and (ΔL) is the change in leg length from foot strike to the middle of stance phase (measured from the ground to the greater trochanter of the femur) [15, 67, 120].

Data Analysis

Once the data gathering trials were completed, each video for each calibration and each trial for each subject was loaded into the ARGUS™ software. This allowed the digitization of the video data and the computation of 2D coordinates for each of the five biomechanical markers. This involved advancing the video to the first frame of contact with the force plate - the first non-zero value on the force plate data was coordinated with the video frame where the foot came in contact with the force plate. From there, each video was iterated five times, with one time for each of the biomechanical markers. The ARGUS™ software, like its MatLab predecessor, does have allowances for the automatic tracking of markers. This automatic tracking needs to be constantly monitored, and can be inaccurate if there is not enough contrast between the marker and the background.

The digitization process resulted in the ARGUS™ program generating a CSV file for each video, containing five paired columns of data. Each column pair represented the X and Y coordinates (respectively) for a biomechanical marker. These coordinates represented the 2D location of each marker, with the values being listed in pixels. Distances between each of the points could then easily be converted into meters by dividing the coordinates into the calibration constant for that day's trial.

The kinematic and coordinate data were smoothed with a fourth-order Butterworth filter, with the cutoff frequency set to approximately 8 hertz [61]. Once this was completed, the calculation of K_{leg}, $K_{Leg-Torsion}$, K_{knee}, and K_{Ankle} could be calculated as described in equations 2, 3, 4, and 5, respectively (see Appendix A for calculations).

The arrays of values for GRF and leg length were iterated through, returning both maximum and minimum values for each variable. This was performed for both control and experimental groups. With each variable that was needed, a new program was written to iterate through a specific group of data. This resulted in data files for GRF and control hop, ACLr hop; as well as files containing leg length for the same groups. These returned variables were then used in the spring-mass calculation of K_{Leg} (see equation 2) for hopping trials

Spring-Mass Calculations

Once the data were smoothed with the Butterworth filter, leg length data were derived by calculating the distance between X and Y coordinates for point 1 and point 5 (head of 5^{th} ray of the foot/distal end of the 5^{th} metatarsal and iliac crest, respectively) on each subject's captured video data. Following this, Spring-Mass K_{Leg} values for hopping were calculated programmatically using a custom Python program by isolating the peak force values and peak changes in leg length for each trial.

Torque Calculations

Torque calculations of K_{Leg} first required the calculation of K_{Knee}, as seen in equation 3. As with the Spring-Mass calculations, the torque calculations were also performed using the aforementioned Python program. In this case, equations 4 and 5 were calculated first (yielding K_{Knee} and K_{Ankle}), and then the results of those calculations were combined with the subjects' mass data, to calculate LSS based on knee joint torque.

Statistical Design

Non-parametric tests were used to analyze the data due to the change in sample size, the presence of outliers that could not be removed, and the lack of normality in the data (sample size and outliers). If significance was found, the Holm-Bonferroni correction was used to adjust the overall p level of 0.05 to account for the statistical error introduced by making multiple comparisons.

The non-parametric Wilcoxon Signed-Rank test - the equivalent of the paired-samples t-test - was used for within group comparisons between the ACLr leg and unaffected leg of the ACL. The Mann-Whitney tests - which is the equivalent to independent t-test - were used for all independent comparison between the ACLr and control groups.

Also, Cohen's effect size was calculated across all comparisons in this study. Cohen's effect size is a quantitative measure of the magnitude of the experimental effect, while statistical significance is determined by p value. Unlike significance tests, the effect size is independent of sample size[121]. Although the p values can represent whether an effect exists, the p value does not reveal the size of the effect and it is possible to have a non-significant p value but considerable effect sizes.

$$d = \frac{M_2 - M_1}{S_{pooled}} \qquad \textbf{Equation 7}$$

Equation 7 shows the formula used for calculating Cohen's d, with M being the mean of each group, and S_{pooled} being the pooled standard deviation of the two groups.

Chapter 4

Results

Hypothesis #1

To determine if ACLr subjects had a quantifiable change in LSS during hopping compared to the control (non-ACL) subjects, as measured by changes K_{leg} Spring-Mass, K_{leg} Joint Torsion, knee and ankle stiffness (K_{Knee} and K_{Ankle}, respectively).

Figure 4 shows the results, for all subjects in this study, of the Mann-Whitney U test of K_{Leg} for both the Spring Mass calculation method (U = 106, p = 0.004) and the Joint Torsional Stiffness calculation method (U = 189, p = 0.44). From these findings, there is a statistical significance between the control and experimental groups for the Spring-Mass method of calculation, which is supported by a large effect size (Cohen's d = 3.14).

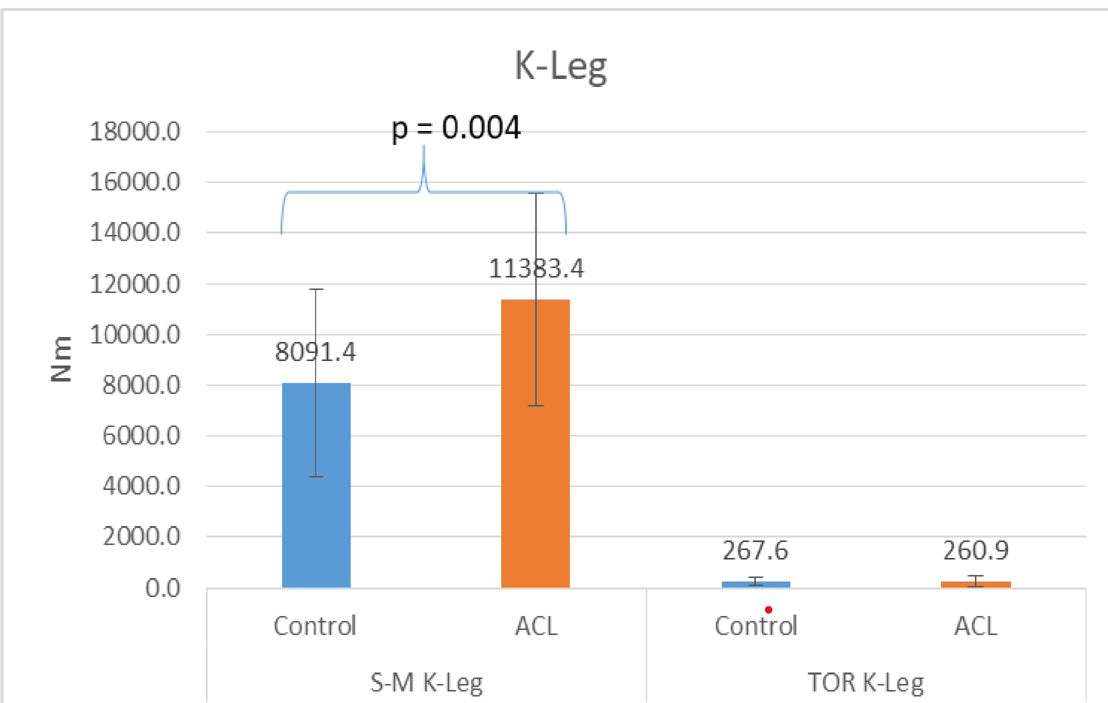

Figure 4. Mean and SD for Spring Mass and Torsional K-Leg values

The results of the Mann-Whitney U test comparing hopping frequencies of the control and experimental groups are shown in Figure 5 (U = 146 Z = -2.27). The Cohen's effect size for this test showed a relatively small effect size (d = 0.26).

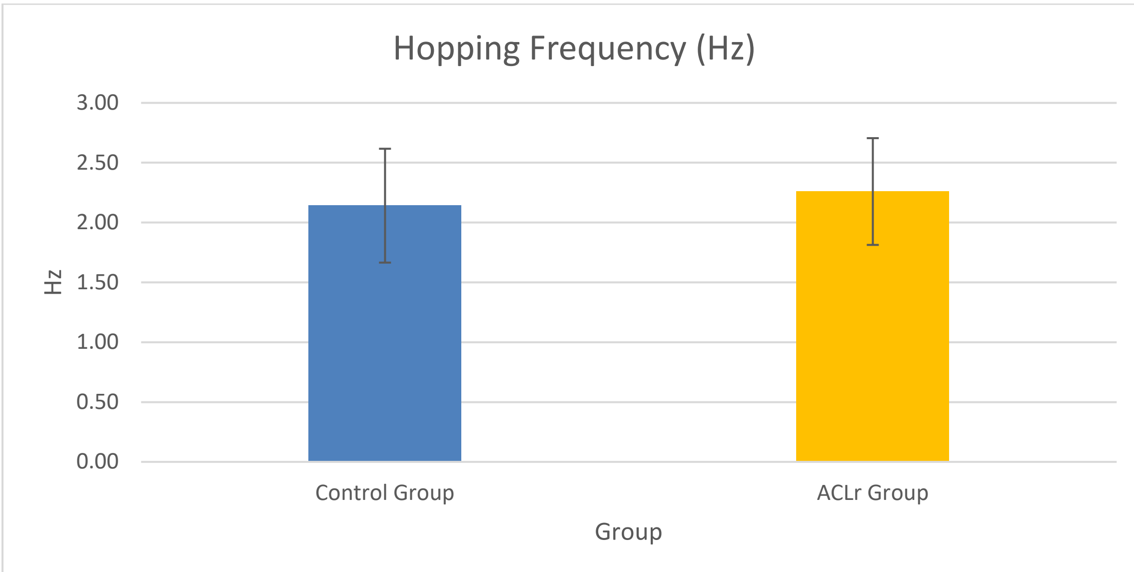

Figure 5. Mean Hopping Frequencies

Figures 6 shows the results, for all legs in this study, of the Mann-Whitney U test of K_{Knee} (U = 178; p = 0.29) and K_{Ankle} (U = 165; p = 0.17), showing no significant difference between groups. Cohen's effect calculations showed small to medium effects for the K_{Knee}, (d = 0.383) but moderate effect size for the K_{Ankle}, (d = 0.541).

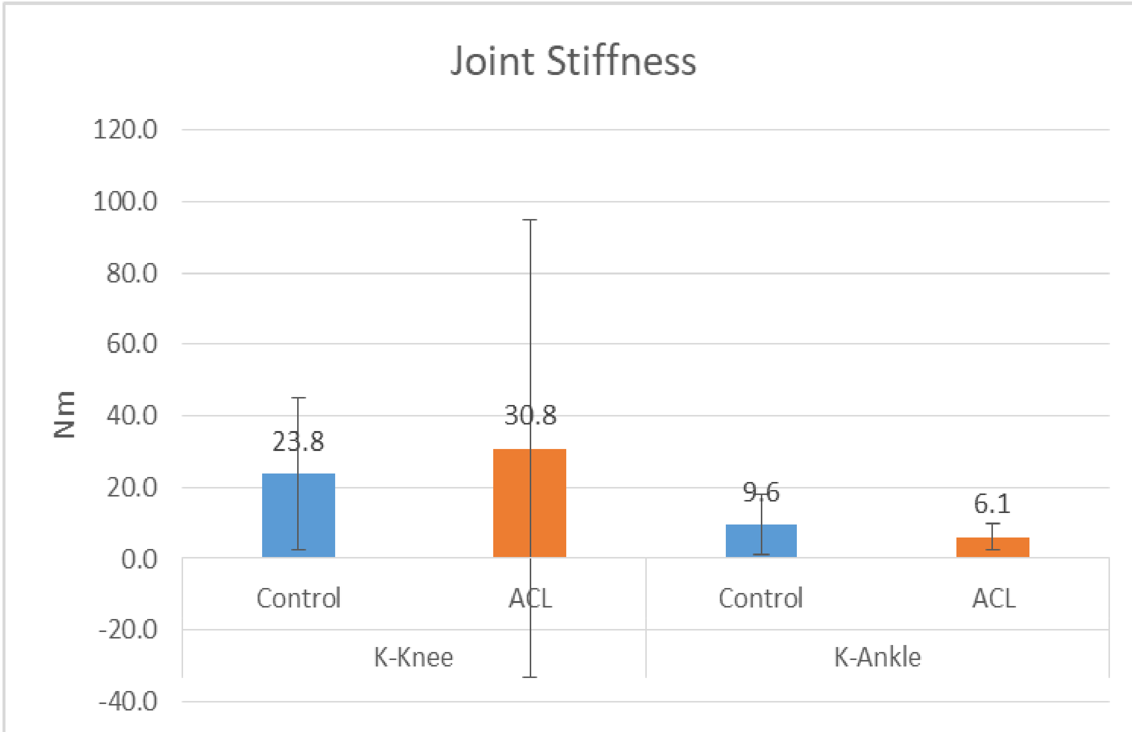

Figure 6. Mean and SD for K-Knee and K-Ankle torsional stiffness values

Hypothesis #2

To determine if there are differences in LSS calculation methods, when comparing vertical center of mass displacement (Spring-Mass) with knee torsion stiffness calculation methods, as well as between the control/non-ACL and experimental/ACL groups.

Figure 7 shows the K_{Leg} values for all the legs computed using both methods. The Spring-Mass method (X-axis) yielded higher values since it represents the overall movement of the center of mass, while the torque method (Y-axis) utilizes the moment of inertia of the thigh segment of the leg.

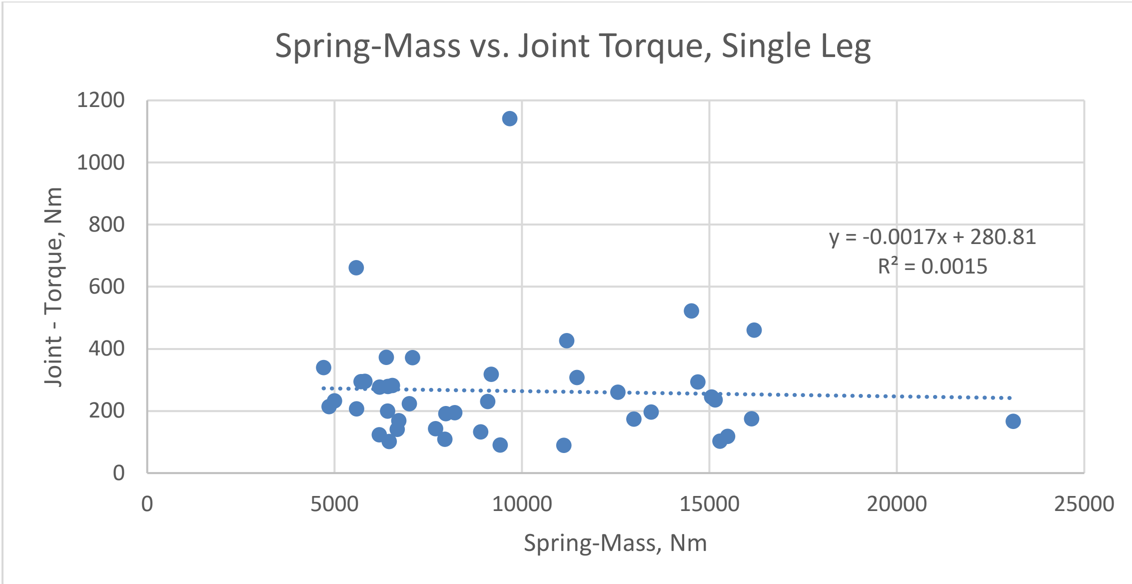

Figure 7. K-Leg Values for all subjects for both computational methods

Figure 8 shows the results of the non-parametric Wilcoxon Signed Rank comparison for all the legs and (N=42) between computational methods. There were significant differences between computational methods across all the subjects' legs (Z = 5.65, p = 0.000), and a large effect size between the 2 calculation methods (Cohen's d = 3.14).

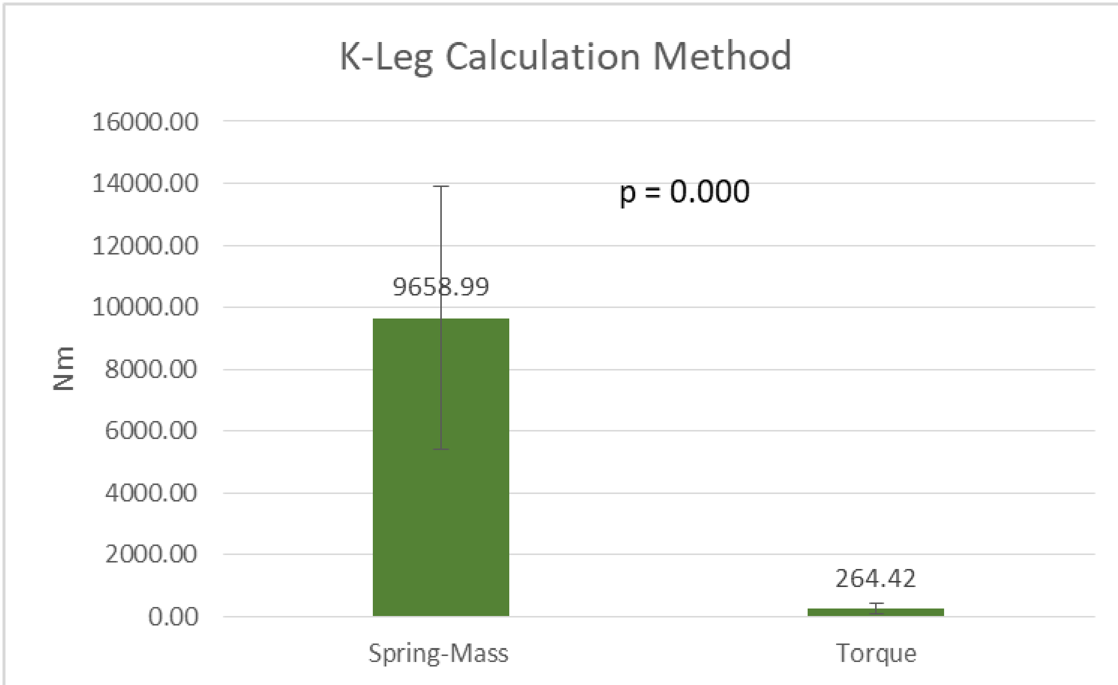

Figure 8. Means and SD for both K-Leg methods

Figure 9 show the results of the Wilcoxon Signed Rank comparison between the two computational methods on only the ACLr leg values as also being significant ($p = 0.005$, Cohen's $d = 4.88$).

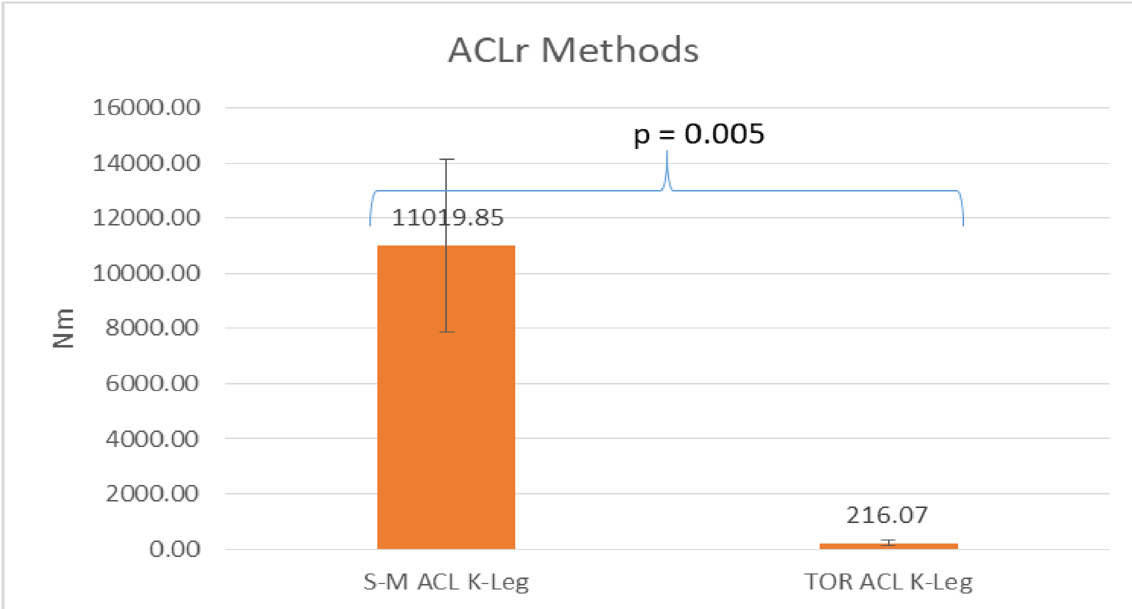

Figure 9. Means and SD for both ACL K-Leg Methods

Hypothesis #3

To determine if ACLr subjects had a quantifiable change in LSS in the affected leg as opposed to the unaffected leg, within the experimental group, as measured by changes in knee and/or ankle stiffness (K_{Knee} and K_{Ankle}, respectively).

As shown in Figures 10 and 11, Wilcoxon Signed-Rank test results for ACL Leg vs Non-ACL leg for experimental group subjects were similar, but not significant in either calculation method (Spring-Mass $p = 0.20$, $Z = -1.27$; torque calculation $p = 0.96$, $Z = -0.05$). The effect sizes for these calculations were: Spring-Mass effect size = 0.170; torque effect size = 0.405.

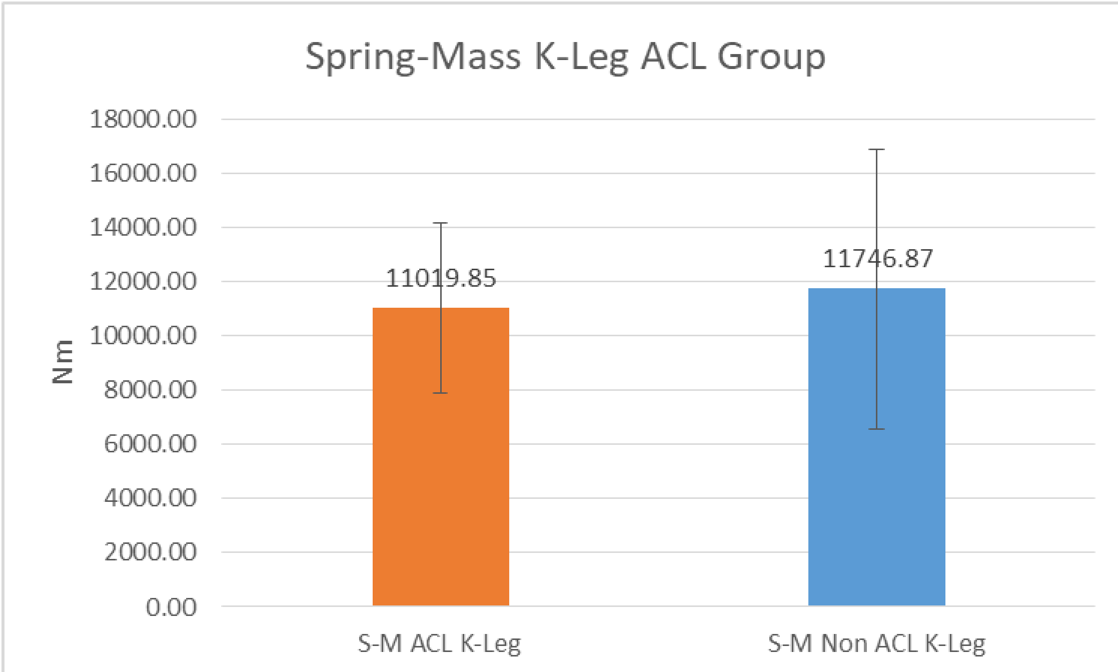

Figure 10. Experimental group Spring-Mass comparison of K-Leg

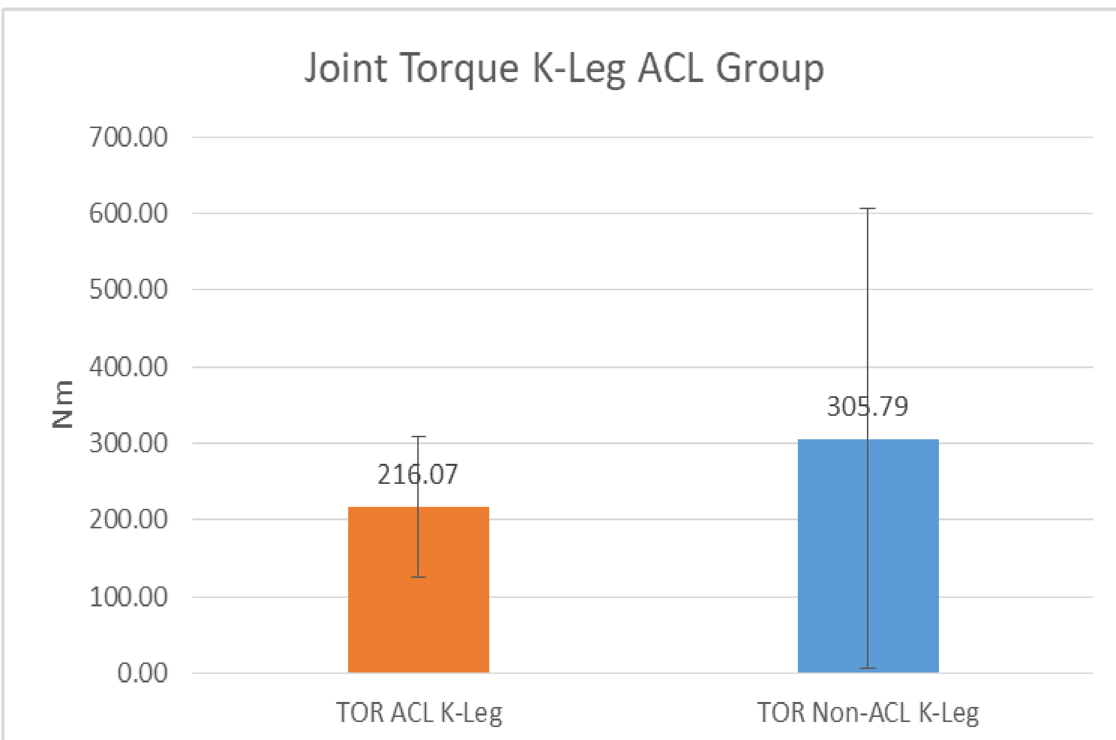

Figure 11. Experimental group torsional comparison of K-Leg

The Non Parametric Wilcoxon signed rank test showed no significant difference for ACL Leg vs Non-ACL leg in the experimental group for K_{Knee} in Figure 12 ($p = 0.80$, $Z = -0.26$) and K_{Ankle} in Figure 13 ($p = 0.39$, $Z = -0.87$), with an effect size of $K_{Knee} = 0.43$ and a $K_{Ankle} = 0.29$.

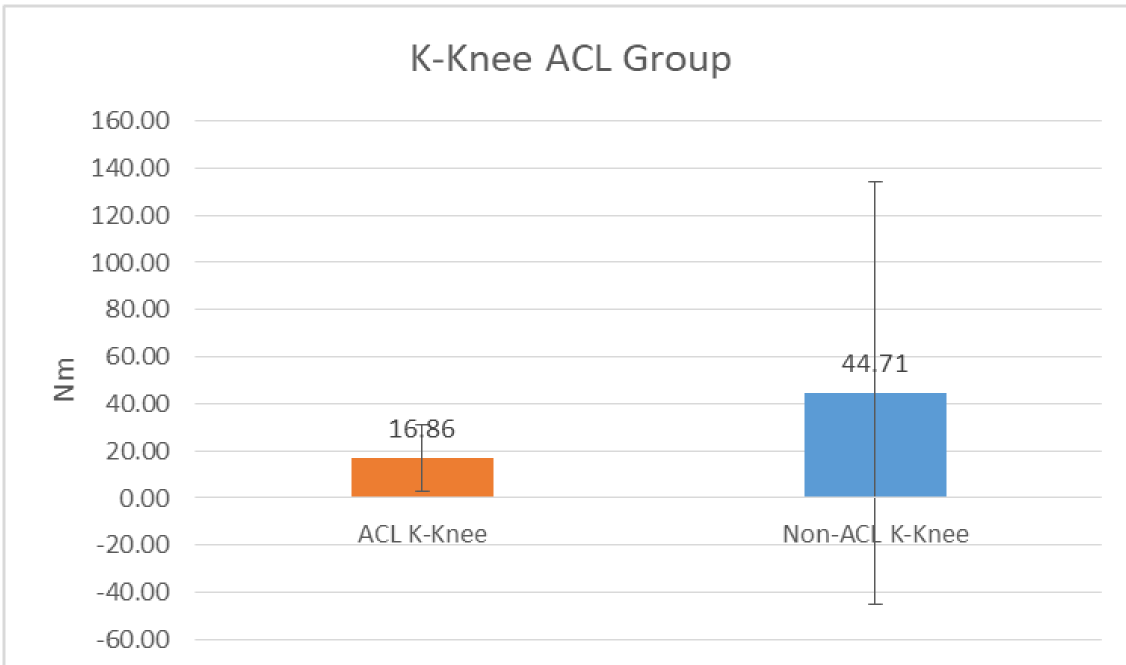

Figure 12. Means and SD for K-Knee

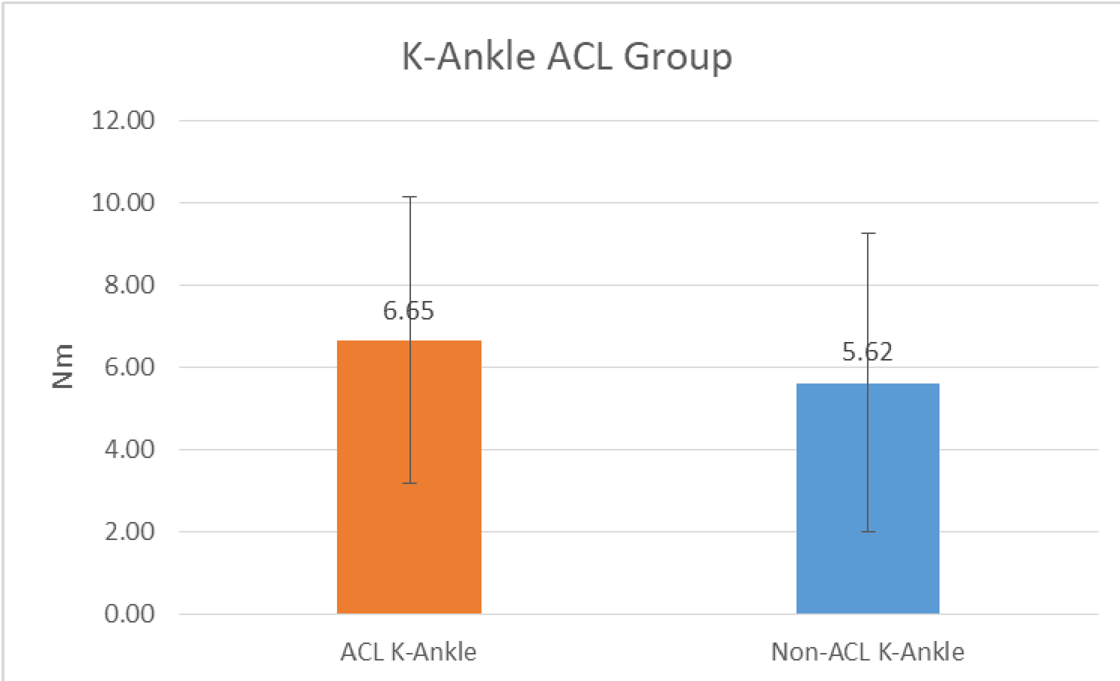

Figure 13. Means and SD for K-Ankle

Chapter 5

Discussion/Conclusions

In this study, hopping trials consisted of a single, 10 second period of continuous hopping for each leg. The values for each subject's trial were combined, and the mean was reported. This 10 second period of data collection resulted in each subject having between 15 and 22 individual hopping events being recorded, allowing for somewhat more accurate measurement of K_{Leg}, and also resulted in a smaller standard deviation. The subjects were allowed to self-select their hopping frequency, which resulted in an average of approximately 2.2Hz. The variation in actual hop number resulted from their activity at the start of the trial. Some subjects hopped on the forceplates immediately, but others stepped on and then began hopping. By allowing the subjects to self-select their hopping frequency, it provided a normalized LSS that could be compared across groups, preventing alterations to LSS that could be caused by a less than ideal frequency for each subject.

The subjects in this study were a combination of male and female subjects, with female subjects composing over 50% of both groups in the study. Due to the increased Q angle of the female subjects, the female subjects were expected to have a lower LSS than their male counterparts [11, 81, 82]. This was consistent with the established literature, but also shows the effect that ACLr had on their LSS, due to the experimental group having statistically significant higher LSS values, regardless of gender.

This study was based on the assumption that ACL damage and reconstruction will have an effect on the leg spring stiffness of the individual. The extent of this effect would be determined by the presence or absence of an ACLr, the mathematical calculation of LSS, and measured by Cohen's d or the effect size. By calculating the effect size for each test pairing in this study, it was possible to determine the magnitude of the difference between control and experimental groups. The calculation of p value determined if that difference would affect the characteristics of the leg spring[121].

Hypothesis #1

To determine if ACLr subjects had a quantifiable change in LSS (LSS) during hopping compared to the control (non-ACL) subjects, as measured by changes K_{leg} Spring-Mass, K_{leg} Joint Torsion, knee and ankle stiffness (K_{Knee} and K_{Ankle}, respectively).

Because of the variation in precision in calculation methods, the first research hypothesis, dealing with the difference between control and experimental subject groups, can be partially accepted. This is due to the statistical significance between the control and experimental group leg spring stiffness, as calculated by the Spring-Mass method of calculation, Figure 4. The research hypothesis cannot be completely accepted, due to the lack of corroboration of these findings with the joint torque method of calculation. This lack of precision in calculation methods continues, when the joint stiffness of the knee and ankle are brought into view. There is some discussion in the literature as to the importance of the changes in LSS being due to the motion of the ankle [1, 15, 111]. With the changes in firing rates and order in the leg muscles of patients who have sustained an ACLi [5, 32, 40, 41, 88], combined with the changes to both medial and lateral hamstring firing rate in the moments immediately prior to foot fall [5, 41, 88, 122], as well as for the tibialis anterior [92, 93], it seems only prudent to more appropriately evaluate the ankle, as well as the knee, when looking at changes to LSS in the ACLr patient.

The data shown in figure 4 demonstrates that in the Spring-Mass method of calculation, there was an increase in the K_{Leg} values for those subjects with ACLr, when compared to those in the control group. This is supported by existing data that shows that there is an increase in overall K_{Leg} value for both legs, when one leg has an ACLr [27, 90, 123], as well as increased accuracy in LSS calculations using this method [15-17, 19, 20, 63, 78, 99]. Consistent with other studies dealing with alterations to the leg-spring due to ACLr, gait alteration and change in LSS are likely due to a change in neuromuscular control of leg-spring muscular activity [26, 32, 35, 93, 95].

The study data is also useful, due to it being closer to the natural activity of unshod, propulsive activity [112, 124, 125], as the foot strike and toe off portions of the unshod running gait are the same as the hopping activity in this study. Being able to absorb shock from the musculature of the leg, by contracting it and allowing it to act as a spring, stores and returns some of the energy generated from gravitational acceleration[84, 112, 113]. Conventional shod running consists of the first ground contact being that of a heel strike. This sends a shock wave through the kinetic chain and can lead to "confusion" by the proprioceptive components of the nervous system. The confusion comes from mechanical dampening of the shock-absorbing components of the shoe initially causing signals to be sent, indicating that the subject is active on a compliant surface, leading to a stiffening of the Leg-Spring. Once the limits of the shoe's shock-absorbing components are reached, subsequent signals are received, analogous to that of the subject being active on a non-compliant surface. The leg is automatically changing its K_{Leg} based on the compliance of the surface being walked, run, or hopped on [15, 111]. If a cushioning and supportive shoe deadens the initial force of impact, due to the initial perception of a compliant surface, once the cushioning limit of the shoe is reached, the stiffer limb now has to deal with the entire force of the impact. This leads to the limb being stiffer than the surface would dictate to an unshod subject[18]. The act of hopping barefoot, with the tactile surface awareness and mid-foot to fore-foot landing pattern, allows for a more natural loading of the kinetic chain.

One potential reason for the increase in stiffness is due to a loss of proprioceptive input, associated with the reconstruction of the torn ACL. Standard surgical procedure for ACL reconstruction, regardless of graft type, involves removing all or nearly all remnants of the torn native ligament and then replacing it with the graft. Following the removal of these structures, the replacement ligament is grafted into place, and secured with the appropriate hardware. Any re-innervation or revascularization of the graft is largely due to serendipity, with little attention paid to the intentional reconnection of neural or vascular structures. This lack of re-innervation and revascularization leads to a significant loss in proprioceptive information being fed into the control systems of the kinetic chain [24, 26, 32, 37, 38, 69, 70, 88, 89, 104, 123, 126].

As stated in the literature, when the ACL is torn, muscular structures such as the hamstrings prevent increased anterior translation of the knee on the femur, with the lateral hamstrings firing at an increased rate to prevent increased rotation of the knee joint. This increase and alteration in muscle firing rate is compensation for the loss of stability that is normally provided by the ACL. Muscles of the lower leg aid in this process by stabilizing the rotatory inputs from the foot hitting the ground. This muscular decrease in lower leg rotation accounts for some of the increases in joint stiffness seen in both the knee and the ankle [27, 29, 34, 71, 87, 88, 94-96]. Once the graft is placed, the lack of innervation and proprioceptive feedback limits the amount of stiffness that can be mitigated by neural feedback loops.

Hypothesis #2

To determine if there are differences in LSS calculation methods, when comparing vertical center of mass displacement (Spring-Mass) with knee torsion stiffness calculation methods, as well as between the control/non-ACL and experimental/ACL groups.

Figure 7 shows the variation in calculation method results that are part of the second hypothesis, with all 21 subjects shown in both calculation methods. The points outside the main cluster on the torque K_{Leg} axis in figure 7 result from two different subjects, both of which have higher than normal K_{knee} values, as a result of high angular velocities (5.7 and 6.7 rad/sec, respectively). What figure 7 also shows is that there may be considerable “repetition without repetition”, which is made apparent when looking at a specific joint for LSS calculations. The end result of the hopping activity may be similar, looking at the entirety of the lower extremity. When a single joint is considered, that joints activity could vary a great deal, while the action of the lower extremity remains consistent when viewed as a whole.

A point to be considered, with regard to the torque calculation not differentiating between groups, is the variables going into the calculation itself. This method of LSS calculation considers the change in knee angle and knee angular velocity, as well as the moment of inertia of the thigh. Since the ACL's function, by physiologic definition, is to prevent anterior translation of the tibia on the femur, then ACLr will most likely have an effect on the motion and moment of inertia of the tibia and fibula, and not on that of the femur. So, for there to be a differentiation between the control and experimental group in this study, a torque calculation would need to involve the leg segment of the lower extremity, instead of the thigh segment.

Figure 8 shows the statistically significant difference and large effect size between the two methods of calculation. When the calculation methods are compared against each other using just the experimental group subjects (Figure 9), the statistical significance remains with an increase in overall effect size, resulting in a more pronounced difference between calculation methods. This is mainly due to the discovery of a "fudge factor" having apparently been employed in the reference study, to allow for the results of the torque calculation to be compared to spring-mass calculations of other studies.

While the scale of the results from both methods of calculation are different in their absolute value, there is no apparent agreement between the two methods used in this study. LSS can be calculated in several different ways [1, 15, 30, 31, 45, 80], with similar, but not identical, resultant values. The different scales in resultant values is due to the variety of variables used in each method of calculation. The spring-mass method of calculation has been used in several studies [15-17, 19, 20, 63, 78, 99], and has been shown to be very accurate and repeatable [45, 80]. The joint torque method of calculation has also been shown to be reasonably accurate and repeatable [31, 68, 127], but only when compared to other studies using the same method of calculation. This method of calculation was also not found to have been used to differentiate groups, or in ACLr subjects.

Additional causes for the discrepancy between the two methods of calculation are likely due to how the data was collected. The spring-mass method of calculation, using the parameters of change in leg length and ground reaction force, can be very accurate and consistent across varying hopping frequencies. This is due to the change in leg length being measured in two-dimensions [1, 14, 15, 30, 61, 106, 107, 109]. When joint stiffness is used as the basis for calculation, as in the torque method used here, other sources of error come into play. The addition of using a velocity variable can add experimental error to the calculations. Also, two-dimensional coordinate date are likely not the most appropriate form of data collection [1, 31]. This is due to the vectors used for measuring the change in angle and angular velocity actually exist in three-dimensional space. The knee joint is also active in three-dimensions as it flexes and extends as part of the motion of the leg. As the knee moves through its range of motion, the ACL is one of the main structures limiting the anterior translation and excessive rotation of the knee, which explains why the classic mechanism of injury for an ACL is a “plant and twist” motion of the body. In the presence of an ACLr, significant changes in the muscular activity occur in these patients [5, 32, 40, 41, 88], which are likely not able to be quantified in two-dimensional data gathering, since they are compensating for rotational forces along the long axis of the leg segments.

Another confounding issue with the joint torque calculations of LSS involves the difficulty of establishing a true joint center and axis of rotation [1]. The multiple degrees of freedom in the knee joint, and this accessory motion, have an effect on the outcome of the measurements of knee motion and the accurate calculation of the moment of inertia of the segment used. Another source of error is the fact that while the change in leg length may be consistent from hop to hop, the actual activity of each joint for each hop may be different [43, 46, 47, 57, 58].

The ability to distinguish between the control and ACLr groups may also be due to the fact that there is still not consensus on which joint has the biggest role to play in the stiffness of the Leg-Spring [2, 15, 32, 111]. Some studies claim that the knee is most important, while others cite the ankle as the critical joint. Since there is no statistical difference between groups in either the ankle or knee calculations, it would follow that there might be an alternate explanation, involving either the hip joint, or more likely, the multiple joints of the foot.

Hypothesis #3

To determine if ACLr subjects had a quantifiable change in LSS in the affected leg as opposed to the unaffected leg, within the experimental group, as measured by changes in knee and/or ankle stiffness (K_{Knee} and K_{Ankle}, respectively).

Figures 10 - 13 display the difference between legs in the experimental group, comparing the ACLr leg to the unaffected leg in each subject, which was the third hypothesis in this study. The lack of statistical differences in either calculation method suggests that the LSS calculation methods used may not be sensitive enough to determine a difference between the affected and the unaffected leg in the experimental group.

In this study, all subjects were college students between the ages of 18 and 25, and all but one of the ACLr subjects were Division 1 or Division 2 college athletes. As such, they have almost all had extensive rehabilitation following their respective injuries, as well as the fact that they were all in competitive shape prior to sustaining their injuries. An ACL injury has not been a career-ending injury for several decades, so all of the subjects that were competitive athletes prior to their injury returned to competitive play following said injury. So their desire to return to play may have aided in their ability to achieve normalcy and parody with the unaffected leg.

Conclusions

In summary, the presence of an ACLr in a subject will not visibly affect their motion pattern in hopping. Their leg will be stiffer, with most of the increased stiffness coming from the contributions of the ankle joint. This increase in stiffness will be relatively uniform across all parts of the hopping activity and will be reflected bilaterally in the unaffected leg. The bilateral alteration to the hopping activity is most likely due to neural adaptations that have occurred, as a protective measure, to limit the motion of the knee and to promote stability and symmetry in the lower extremity. If LSS is the evaluation metric, it should be calculated using the spring-mass method for consistency and repeatability. If the joint torque method is to be used, three-dimensional kinematic analysis should be employed for a more appropriate representations of the motions of the knee and ankle joints, as the two-dimensional kinematic analysis does not completely describe the leg-spring system. The two methods of calculation should also be considered independent of each other, and comparison between methods should be avoided, to prevent misleading data from being presented.

Given the small effect size present in the ankle joint, and the moderate effect size present in the knee, increased precision in data collection and calculation should yield statistically significant results. As the study was limited by a small sample size, future replications of this study should involve a much larger sample size. Additionally, additional joint markers should be used, to more accurately calculate joint centers for the three-dimensional kinematic analysis.

Appendix A

The purpose of this appendix is to show the calculation procedure for both the Spring-Mass method of calculating K_{Leg}, as well as the torque method of calculating K_{Leg}, with the calculation of both K_{Knee} and K_{Ankle}. The following data will be used for example purposes.

Subject	GRF (n)	Leg Length Change (m)	Mass (kg)	Thigh Length (*l*,m)	Foot Length (m)	Change in Knee Angle (Θ, rad)	Knee Angular Velocity (ω, rad/sec)	Change in Ankle Angle (Θ, rad)	Ankle Angular Velocity (ω, rad/sec)
Control	901.0	0.191	68.18	0.4176	0.1068	2.580	4.19	2.24	6.34
Non-ACL	1428.9	0.089	79.55	0.4516	0.1412	2.666	2.67	2.36	5.41
ACLr	1233.0	0.150	90.91	0.3016	0.1226	2.664	2.63	2.22	5.12

Control Spring-Mass (K_{Leg}) calculation:

$$K_{Leg} = F/\Delta L$$

$$K_{Leg} = 901.0/0.191$$

$$K_{Leg} = 4705.7 \text{ Nm}$$

Control K_{Knee} calculation:

$$I = \text{ } Thigh\ Length^2 \times Mass$$

$$I = \text{ } 0.4176^2 \times 68.18$$

$$I = 11.89$$

$$K_{knee} = I(\Delta\omega^2 \div \Delta\theta^2)$$

$$K_{Knee} = 11.89\ (4.19^2 \div 2.580^2)$$

$$K_{Knee} = 31.51 \text{ Nm}$$

Control K_{Leg} calculation:

$$K_{leg} = K_{knee} \div l^2 sin\Delta\theta$$

$$K_{Leg} = 31.55 \div 0.4176^2 sin2.580$$

$$K_{Leg} = 31.52 \div 0.174 sin2.580$$

$$K_{Leg} = 340.0 \text{ Nm}$$

Control K_{Ankle} Calculation:

$$I = \text{ } Foot\ Length^2 \times Mass$$

$$I = \text{ } 0.1068^2 \times 68.18$$

$$I = 0.78$$

$$K_{ankle} = I(\Delta\omega^2 \div \Delta\theta^2)$$

$$K_{Ankle} = 0.78(6.34^2 \div 2.24^2)$$

$$K_{Ankle} = 6.01\text{Nm}$$

Non-ACL Spring-Mass (K_{Leg}) calculation:

$$K_{Leg} = F/\Delta L$$

$$K_{Leg} = 1428.9/0.089$$

$$K_{Leg} = 16127.3 \text{ Nm}$$

Non-ACL K_{Knee} calculation:

$$I = \ Thigh\ Length^2 \times Mass$$

$$I = \ 0.4516^2 \times 79.55$$

$$I = 16.22$$

$$K_{knee} = I(\Delta\omega^2 \div \Delta\theta^2)$$

$$K_{Knee} = 16.22\ (2.67^2 \div 2.666^2)$$

$$K_{Knee} = 16.27 \text{ Nm}$$

Non-ACL K_{Leg} calculation:

$$K_{leg} = K_{knee} \div l^2 sin\Delta\theta$$

$$K_{Leg} = 16.27 \div 0.4516^2 sin 2.666$$

$$K_{Leg} = 16.27 \div 0.204 sin 2.666$$

$$K_{Leg} = 175.3 \text{ Nm}$$

Non-ACL K_{Ankle} Calculation:

$$I = Foot\ Length^2 \times Mass$$

$$I = 0.1412^2 \times 79.55$$

$$I = 1.59$$

$$K_{ankle} = I(\Delta\omega^2 \div \Delta\theta^2)$$

$$K_{Ankle} = 1.59(5.41^2 \div 2.36^2)$$

$$K_{Ankle} = 2.03 \text{ Nm}$$

ACL Spring-Mass (K_{Leg}) calculation:

$$K_{Leg} = F/\Delta L$$

$$K_{Leg} = 1233.0/0.150$$

$$K_{Leg} = 8220.0 \text{ Nm}$$

ACL K_{Knee} calculation:

$$I = \ Thigh\ Length^2 \times Mass$$

$$I = \ 0.3016^2 \times 90.91$$

$$I = 8.27$$

$$K_{knee} = I(\Delta\omega^2 \div \Delta\theta^2)$$

$$K_{Knee} = 8.27\ (2.63^2 \div 2.664^2)$$

$$K_{Knee} = 8.06 \text{ Nm}$$

ACL K_{Leg} calculation:

$$K_{leg} \ = K_{knee} \div l^2 sin\Delta\theta$$

$$K_{Leg} = 8.06 \div 0.3016^2 sin2.664$$

$$K_{Leg} = 8.06 \div 0.091 sin2.664$$

$$K_{Leg} = 194.4 \text{ Nm}$$

ACL K_{Ankle} Calculation:

$$I = \ Foot\ Length^2 \times Mass$$

$$I = \ 0.1226^2 \times 90.91$$

$$I = 1.37$$

$$K_{ankle} = I(\Delta\omega^2 \div \Delta\theta^2)$$

$$K_{Ankle} = 1.37(5.12^2 \div 2.22^2)$$

$$K_{Ankle} = 7.28 \text{ Nm}$$

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Curriculum Vitae
David K. Wolfe

Formal Education:

1992	East High School Akron, Ohio
1998	University of Akron Akron, Ohio B.S. Sports Medicine Education
1999	University of Akron Akron, Ohio B.S. Biology
2004	University of Louisville Louisville, Kentucky M.S. Physiology and Biophysics
2020	Indiana University, Purdue University, Indianapolis Indianapolis, Indiana PhD, Health and Rehabilitation Science

Professional Experience:

American Medical Response
Akron, Ohio

02/28/00 – 09/29/00 Position: Calltaker/Dispatcher, Communications

Responsible for the call intake, scheduling, referral and dispatching of emergent, non-emergent and wheelchair ambulance activity.

WIL Research Laboratories, Inc.
Ashland, Ohio

10/02/00 – 06/30/02 Position: Biologist 1, Inhalation Toxocology

Responsible for conducting acute, subchronic, reproductive, developmental, chronic and special inhalation toxicology studies. Functions include, but are not limited to, test atmosphere generation and analysis, operation and maintenance of all related equipment, operation of the LabVIEW® for Windows Data Acquisition System, responding to Quality Assurance Unit audits, preparation and review of methods, tables and summaries of reports on inhalation studies. Conducting detailed clinical observations on animals on inhalation studies. Collection of all test atmosphere, generation and analysis, chamber environmental and equipment maintenance data related to the conduct of inhalation toxicology studies in accordance with study protocols, standard operating procedures, Good Laboratory Practice standards and appropriate federal and international testing guidelines. Supply requisition. Inhalation department Safety Committee representative. Inhalation Technician of the Year. Other duties as assigned.

University of Louisville
Louisville, Kentucky

11/01/02 – 10/01/03 Position: Research Technician Physiology and Biophysics

Responsible for the setup, conduct and data collection of experiments and experiment related equipment and activities. Care and maintenance of laboratory equipment. Ordering and stocking of laboratory supplies. Surgical procedures associated with experimental protocol. Other duties as assigned.

University of Louisville
Louisville, Kentucky

9/15/03 – 12/20/03 Position: Physiology Tutor

Responsible for tutoring Dental students in Dental Physiology courses, through the University of Louisville School of Dentistry.

Quest Outdoors at the Summit
Louisville, Kentucky

11/01/03 – 10/01/04 Position: Sales Associate

Responsible for customer service; proper fitting, placement, operation, setup, maintenance, and installation of outdoor adventure equipment and related gear. Top Sales Associate, first six months. Other duties as assigned.

Indiana University Southeast
New Albany, Indiana

08/15/04 – 07/31/07 Position: Adjunct Faculty

Responsible for the instruction of Basic Human Physiology, P215, sections 23216, 23217, 22261, 12210, 1926, 1907, and 6550 including associated laboratory sections. Additional duties associated with college level classroom and laboratory instruction.

Spalding University
Louisville, Kentucky

04/25/05 – 07/31/07 Position: Adjunct Faculty

Responsible for the instruction of Human Anatomy and Physiology, sections 251 and 253 (lecture), sections 252 and 254 (laboratory, two sections each), Living Systems Development, Biology section 107 and Fundamentals of Biology 2, sections 102 and 104 (lecture and laboratory, respectively). Additional duties associated with college level classroom and laboratory instruction.

Bellarmine University
Louisville, Kentucky

12/01/05 – 07/31/07 Position: Adjunct Faculty

Responsible for the instruction of Introduction to Life Sciences, Lecture (Biology 115 KM) and associated Laboratory section (Biology 115L 01). Additional duties associated with college level classroom and laboratory instruction.

Brownings Brewery and Restaurant
Louisville, Kentucky

10/05/05 – 12/15/06 Position: Assistant Brew Master, Bartender

Responsible for and assisting in the formulation and production of beer for restaurant and off site use with capacity extending to include 15 barrels production (2445 Liters), customer service and customer relations.

STAR Physical Therapy
Charlestown, Indiana

08/15/06 – 07/31/07 Position: Head Athletic Trainer, Charlestown High School

Responsible for all on-site athletic training activity for all Varsity, Junior Varsity, Freshman and Junior High athletic teams, as well as all away athletic training activity for Varsity Football. Teams included: Football, Men's and Women's Basketball, Men's and Women's Cross Country, Men's and Women's Track and Field, Baseball, Basketball, Wrestling, Swimming, Volleyball, Golf, Men's and Women's Tennis, and Soccer. Indiana State Athletic Training License 36001352A.

University of Indianapolis
Indianapolis, Indiana

08/01/07 – 08/15/10 Position: Assistant Athletic Trainer, Approved Clinical Instructor

Athletic Training Assignments: Varsity Men's Soccer and Baseball, traveling Athletic Trainer with Women's Soccer. Other duties include supervision of Athletic Training students and candidates, as well as Athletic Training Room instruction, observation, and evaluation. Indiana Athletic Training Association (IATA) Quiz Bowl Coach for the University of Indianapolis. Indiana State Athletic Training License 36001352A. RUIndy Fit Team Captain. Athletic Trainer for 2009 NCAA Regional Volleyball Tournament.

University of Indianapolis
Indianapolis, Indiana

01/14/09 – 08/15/10 Position: Adjunct Faculty, Department of Kinesiology

Department of Kinesiology, Classroom instruction for Principles and Practices in Exercise Science, KINS 245-01 and 02.

National Soccer Coaches Association of America
Kansas City, Kansas

01/13/09 – 01/30/10 Position: Athletic Trainer

Athletic Training coverage of national convention in St. Louis, Missouri in 2009, and 2010 in Philadelphia, Pennsylvania.

The University of Indianapolis
Indianapolis, Indiana

07/16/10 – 09/20/19 Position: Instructor of Biology, Approved Clinical Instructor

Classroom and Laboratory instruction for: BIOL – 103 (Basic Human Anatomy), BIOL – 104 (Basic Human Physiology), FYS – 130 (First Year Seminar; "Are you going to eat that?"), BIOL – 200 (Medical and Scientific Terminology), BIOL – 305 (Functional Human Anatomy), BIOL – 330/BIOL-505 (Mammalian Physiology; Undergraduate / Graduate, respectively), BIOL – 503 (Clinical Human Anatomy). BIOL – 104 Course Director, Undergraduate Research Faculty Mentor. BIOL – 509 Clinical Pathophysiology guest lecturer. At various times: Interim Faculty Senate and Committee member, Summer School Task Force, University of Indianapolis Critical Response Team. Lilly Science Hall Building Safety Warden. University of Indianapolis Symphonic Wind Ensemble, University of Indianapolis Hound Sound Pep-Band.

Crossroads of America Council, Boy Scouts of America
Indianapolis, Indiana

10/15/19 – 07/13/20 Position: District Executive, Sugar Creek District

The District Executive is an entry-level position that has responsibilities that are broad and varied. Duties include promoting, supervising, and working through adult volunteers including parents and community leaders.

Different aspects of the position include: Sales. The executive is responsible through volunteers for extending Scout programs to schools, religious, civic, fraternal, educational, or other community-based organizations. Service. The executive provides quality service through timely communication, regular meetings, training events and activities. Fundraising. Working with volunteers, executives recruit leadership for the Friends of Scouting and fundraising campaign efforts to meet the financial needs of the council. Administration. The District Executive administers the Scouting program in the assigned district service area. Public Relations. The District Executives must be a good role model and must recognize the importance of good working relationships with other professionals and with volunteers. Scouting depends on community support and acceptance. Professional leaders must have good communication skills and be able to tell Scouting's story to the public.

<u>Scholarship:</u>

NATA BOC Certified Athletic Trainer, 1999

The FASEB Journal; Experimental Biology 2004
Washington D.C.; April 17 – 21, 2004
Poster Presenter: *Hemodynamic Effects of Exogenous ATP Delivery*

Research Louisville
Louisville, Kentucky; November 8 – 12, 2004
Poster Presenter: *Hemodynamic Effects of Exogenous ATP Delivery*

“Inside the Game”
Southern Indiana High School Athletics
Volume 5, Issue 18, January 18-24, 2007
STAR Sports Medicine Corner: *What is creatine and who should use it?*

“Inside the Game”
Southern Indiana High School Athletics
Volume 5, Issue 19, January 25-31, 2007

STAR Sports Medicine Corner: *Sports Supplements: Are they worth it?*

NATA Day on Capitol Hill
February 26, 2007

"STAR IMPACT"
STAR Physical Therapy
Injury Prevention/Sports Performance Reference

Indiana Athletic Trainers Association (IATA), 2008
Education Committee Member, University of Indianapolis
"Quiz Bowl" Coach.

Michigan State University, 2009
East Lancing, Michigan
Osteopathic Manipulative Medicine for Athletic Trainers Seminar
Levels 1 and 2

Indiana Interdisciplinary Conference, Spring 2017
"What is growing in your locker room? Engaging undergraduate research fellows in STEMing the stench**."**
Co Author and Faculty Undergraduate Research Mentor.

Indiana College Biology Teachers Association
Annual Meeting, October 2017
"Universal Design in Education; Improving Instruction and Exam Scores"

National Science Teachers Association
Journal of College Science Teach Advisory Board
January 2018 – Current

Anatomage Table User Group
Annual Meeting, July 2019
San Jose, CA

QPR Gatekeeper Training
Suicide Prevention
September, 2020

Grants:

Harris-Manchester College Summer Research Institute, Visiting Fellow, Summer 2012.

IATA Graduate Scholarship, Fall 2010

KBRIN Grant Recipient, Spring 2004

University of Louisville, School of Medicine, Travel Award Recipient, Spring 2004

Honors:

Dean's List

Eagle Scout

Inhalation Toxicology 2002 Technician of the Year